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Appendix H
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Gait Analysis of a New Low Cost Foot Prosthetic for Use in Developing Countries

By

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A thesis submitted to the School of Physical and Health Education in conformity with the requirements for the degree of Master of Science

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Abstract

Before a new foot prosthesis can be introduced to the public, gait testing must be performed to ensure that the gait patterns produced while wearing the foot are acceptable. At present there are no guidelines to determine what level of performance is acceptable. The most common method used to determine the suitability of a new foot prosthesis is to compare its gait pattern to the pattern of an existing prosthetic foot that has been deemed acceptable. Large deviations in the certain kinetic properties, such as the moments produced at the knee joint, can have detrimental affects on the individual wearing the foot.

A new prosthetic foot design, created by the Niagara Prosthetics and Orthotics company (NPO), was brought to Queen's University, for the clinical testing. A comparison study with the Stationary Ankle Flexible Endoskeleton foot (SAFE) was used to determine if there were differences in the gait pattern while wearing the NPO foot and whether these differences would be problematic. Five below knee amputees volunteered to attend two testing sessions at the gait laboratory in Kingston General Hospital. Prior to the first session they were fitted with a SAFE prosthetic foot. Prior to the second visit they were fitted with the new NPO prosthetic foot. All fittings were performed by the same prosthetist. Subjects were allowed a minimum of two days to accommodate to each prosthetic foot. Subjects walked at self-selected speeds across a six meter walkway that had an AMTI force plate mounted in the floor. An optoelectric motion tracking system was used to collect information on a series of 12 markers placed at selected body landmarks. At the end of each testing session the subjects filled out a subjective questionnaire about the performance of the foot tested.

The two prosthetic feet were compared in four main categories: time distance parameters, gait curve patterns, gait curve parameters, and a subjective questionnaire. A two way ANOVA with repeated measures reveled that eight of the variables differed between the feet. The stance ratio was found to be smaller in the NPO. The maximum moment (AP) at the knee joint was lower in the NPO foot. The maximum moment (LM) occurred earlier in the gait cycle for the NPO foot. The NPO produced a larger degree of ankle dorsiflexion. The NPO foot generated a smaller braking impulse and a smaller

propulsive impulse. The NPO had a higher vertical GRF peak and a lower vertical GRF slope suggesting that it absorbs less energy but does it over a longer period of time. Several of these differences were attributed to the lack of adequate cushioning, or shock absorptive properties in the heel of the NPO prosthetic foot. Although several variables were found to be significantly different between the prosthetic feet, no difference in NPO foot gait pattern were considered problematic and the foot was deemed acceptable.

The NPO foot was designed to meet the needs of individuals in developing countries. The high cost of modern foot components, short life expectancy, and lack of skilled prosthetists to fit and maintain complex prosthetics, combined with annual incomes far below those of North American countries, has created a serious need for an inexpensive, uncomplicated, and durable prosthetic foot able to produce adequate gait patterns. The NPO combines many of these attributes with its elegant one piece design produced from a polyethylene compound that is inexpensive, durable, flexible, and can be injection molded. The result is a prosthetic foot that can be produced at an estimated cost of $7-10 each.

Despite some small short comings in performance, when compared to the SAFE foot, the NPO foot produced an adequate gait pattern. The foot's ability to meet the specific needs of a third world market in terms of cost, durability and simplicity, makes the NPO foot a potentially successful alternative to more costly components. Questions as to its acceptability in countries with cultural sensitivities and cultural-specific needs remain unanswered.

Acknowledgments

I think that there are few times that we stop and consider how many people help and support us in the things that we do. So now that I have that opportunity get comfortable, the list of people who deserve my thanks for their help and support during my thesis is long.

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Our gait study was part of a larger study on the NPO foot, conducted in conjunction with people from the clinical mechanics group. Dr. Tim Bryant helped to coordinate the different aspects of the study, and also provided input for our gait study. Amy Tsai, choose the materials for the NPO foot. Amy was a tremendous help through my entire study. She had input into the development of the study protocol, helped solve our

"slipping" problem, and made late night, last minute book deliveries to help me finish my work. Gerry Saunders helped make last minute adjustments to the foot, rounding corners of heels, and toes at a moments notice to make a square foot fit in a round shoe.

Due to confidentiality issues I'm not permitted to print the names of my subjects. But I would like to thank each of them (CL, DK, MA, MM, SG, they know who they are). There was no financial reimbursement available for participation in this study, and the individuals who volunteered their time did so in the hopes that people less fortunate than they may benefit. The feedback that was provided to us from our subjects was valuable to the study and the foot's designer.

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Although I had many people help me with the editing of my work, no one was as patient as Megan. Thank you for patients and support, especially near the end to the project (and for the figure in the methods section).

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1.1 Introduction

Amputation of a lower limb is most commonly performed due to trauma or disease of the limb (Davies, Friz, et al. 1970). The two most often performed amputation procedures are truncation of the femur bone Above the Knee (AK) and truncation of the tibia bone Below the Knee (BK)(Fishman & Watkins 1974)(Davies, Friz, et al. 1970).

Physical loss of the anatomy of the lower limb results in loss of gait function. In BK amputees this loss is due partly to the loss of the articulated ankle joint, the loss of the joints of the foot (including the metatarsal-phalangeal joints), and the loss of the muscles of the anterior and posterior compartments of the shank (Rose & Gamble 1996)(Winter 1991). The purpose of a below knee leg prosthesis is to replace the function lost due to the physical loss of anatomy.

BK prostheses are typically comprised of four major components (Figure 1.1), these are:

1) Socket
2) Pylon
3) Foot prosthetic
4) Couplings

The almost universal standardization of the pylon and coupling components of BK prosthetics allows for the easy interchanging of components. The BK socket that

supports the residual limb (or stump) is normally custom built to the individual's anthropometric specifications (Davies, Friz, et al. 1970)

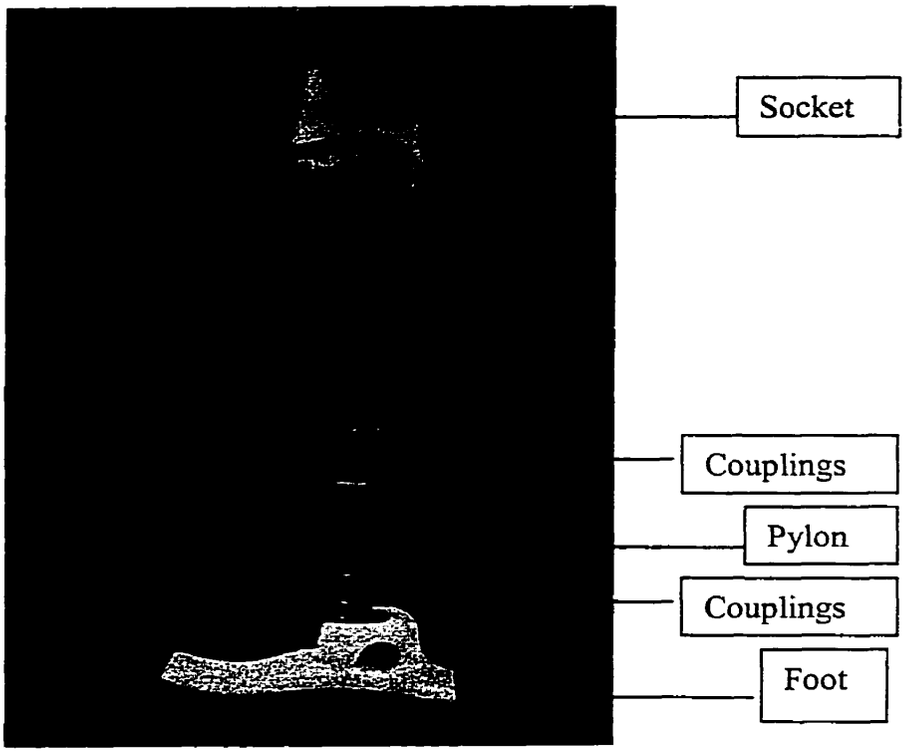


Figure 1.1: Components of a below knee prosthetic limb

Presently there are a large number of commercially available prosthetic feet. All prosthetic feet attempt to return some of the lost gait function, but may use different mechanical principles to do so. Some feet attempt to replace the function lost due to the absence of an articulated ankle joint by providing a mechanical articulating joint, while

others attempt to restore this function using different mechanics. These mechanics include the use of rubberized materials in the heel of a prosthetic foot that compresses on heel strike simulating ankle plantarflexion provided by an articulating joint. Although different mechanical approaches are used, the goal of all foot prosthetics is the same: to provide a device that produces proper gait kinematics without adversely affecting the kinetics of the knee and hip. A prosthetic foot that allows proper gait kinematics, yet produces excessive moments or abnormal gait patterns would render it unusable (Perry 1975). Therefore, when evaluating a foot prosthesis it is necessary to observe both the kinematic and kinetic gait characteristics produced while using the foot (Rose & Gamble 1996).

Research and design in the field of prosthetics is an industry largely driven by the interests of developed countries and therefore considers the needs and prosperity of these developed countries. Research often aims to "optimize" the gait produced by prosthetic feet and the general trend in prosthetics has been towards more intricate multi-piece designs that use sophisticated energy return materials (Bartkus, Colvin, et al. 1994). The result of this trend has been large increases in the cost of modern prosthetics due to: material cost, production cost, and requirement of skilled individuals to fit and maintain these prosthetics (Meanley 1995). New specialized custom built prosthetics designed to allow amputees to participate in sports can cost in excess of five thousand dollars but the cost of normal use prosthetic feet is also on the rise. Prices for many of the most commonly used feet range from $150 - $800 dollars. While the impact of this trend is not as dramatic in countries with developed health care systems, the same cannot be said of underdeveloped countries.

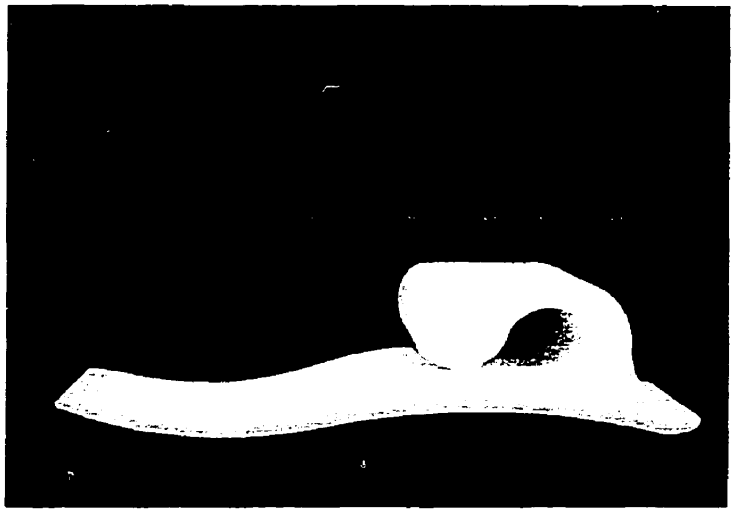


Figure 1.2: The Niagara Prosthetics and Orthotics prosthetic foot design (NPO)

Contrary to this trend of escalating cost, the Niagara Prosthetics and Orthotics Company (St. Catherine's, Ontario) designed a prosthetic foot where low cost and simplicity were the driving principals behind the design. The resulting foot, referred to herein as the Niagara Prosthetics and Orthotics prosthetic foot (NPO) (Figure 1.2), is a simple single piece device designed to attach to a standard prosthetic pylon (Ziolo 1999). The prosthetic foot design was brought to a team of researchers at Queen's University for both mechanical and clinical testing. For the clinical evaluation, a gait study comparing the NPO foot with the SAFE (Stationary Ankle Flexible Endoskeleton) foot was chosen to determine the suitability of the new NPO foot.

The comparison foot, the SAFE foot, belongs to a category of prosthetics known as SACH feet (Solid Ankle Cushioned Heel) that have a compressible (or cushioned) material in the heel of the prosthetic foot (Figure 1.3). A SACH foot was chosen for

comparison because it is relatively inexpensive, available and widely used in underdeveloped countries and considered to be the industry standard for research. A large number of gait patterns and characteristics studies using the SACH feet have been published and it is by far the most used benchmarking foot (Fishman & Watkins 1974)(Davies, Friz, et al. 1970)(Goh, Solomonidis, et al. 1984)

The purpose of the present study was to compare the gait patterns and characteristics produced by the NPO, and SAFE prosthetic feet to observe where differences exist.

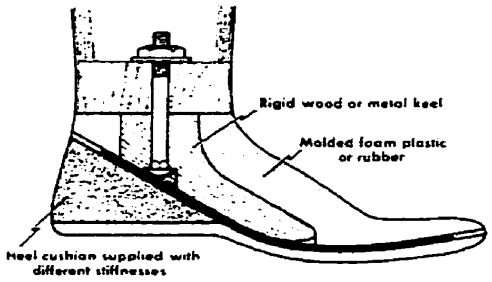


Figure 1.3: The SACH (Solid Ankle Cushioned Heel) prosthetic foot.

1.2 Glossary of Terms

Sound Limb – The non-amputated limb. Also commonly referred to as: contralateral limb or unaffected limb.

Residual Limb – The amputated limb. The anatomy of the lower limb above the line of amputation that remains intact. Also commonly referred to in literature as the stump or affected limb.

NPO – The Niagara Prosthetics and Orthotics prosthetic foot. A low cost single piece prosthetic foot manufactured from polyethylene or nylon.

SAFE – Stationary Ankle Flexible Endoskeleton prosthetic foot design. A prosthetic foot that fits in the SACH category of feet. Similar to the original SACH foot but with the addition of a flexible keel allowing for easier roll over during gait.

SACH – Solid Ankle Cushioned Heel prosthetic foot design. A prosthetic foot with a spongy material in the heel that compresses during heel strike.

BK – Below Knee (amputee or prosthetic). Also referred to in literature as trans-tibial amputee or prosthetic

AK – Above Knee (amputee or prosthetic). Also referred to in literature as trans-femoral amputee or prosthetic.

Cadence – The number of steps per minute, calculated by step/time, express in steps per minute.

Gait Time – Time to complete one gait cycle. The time from heel strike of one foot to the next heel strike of the same foot, expressed in seconds.

Gait Velocity – The rate of forward progression of the body, expressed in meters per second.

Stance Ratio – The ratio of the time spent in stance phase to total time for one gait cycle (gait time / stance time).

Stance time – The total amount of time spent in the stance phase, expressed in seconds.

2.1 Review of Literature

In 1561 Ambroise Pare introduced the first modular prosthetic limb. Unlike its predecessors the modular prosthetic was not a single piece fabrication, but a series of joined components (Sanders 1986). The modular design allowed for the interchange of the foot-ankle component and until recently the choices in this component have been limited. Over the past three decades a large number of new feet have become available (Edelstein 1988). The industry has emphasized research and product development in the high end market (both in terms of function and cost). The number of prosthetic feet affordable to low income amputees has declined (Bartkus, Colvin, et al. 1994)

Research has suggested that abnormalities, such as asymmetrical gait or altered joint moments associated with BK amputations lead to degenerative changes in the back and knees (Perry 1975)(Brouwer, Allard, et al. 1989)(Marks, Palmer, et al. 1978). For example, Lemaire et al. (1990) found evidence of increased osteoarthritus in the contralateral knee joints of elderly BK amputees (Lemaire & Fisher 1994).

However, Rubin (1986) concluded that given similar patient diagnoses the choice of prescribed foot prosthetic varied between prosthetic departments. This suggests that not enough empirical biomechanical data is available to distinguish between new feet based on performance. With more than 500,000 lower limb amputees in North America and an additional 43,000 lower limb amputations performed each year in North America (Hunter, Smith, et al. 1995), it is important to ensure that new prosthetic feet can produce appropriate gait and that performance difference can be shown through empirical means.

2.2 Comparison Study Design

The most common way to test the suitability of a new prosthetic foot is using a comparison study design. In this design a newly introduced prosthetic foot is compared to an existing prosthetic foot whose gait properties have been studied and deemed acceptable. Work by Doane and Holt (1983) used this approach to investigate the gait pattern differences between the SACH and single axis prosthetic feet and they concluded that no significant differences existed between the gait patterns produced by the SACH and single axis feet. Other investigators have used the comparison approach to evaluate the same two feet (SACH and single axis) (Culham, Peat, et al. 1986)(Culham, Peat, et al. 1986) (Goh, Solomonidis, et al. 1984) (Brouwer, Allard, et al. 1989) (Winter & Sienko 1988). While these studies observed a number of different gait parameters, none found significant differences between feet and agreeed with the findings of Doane (1983).

2.3 Choice of the SAFE Foot

By far the most commonly used comparison or "benchmark" prosthetic foot is the SACH foot. The Solid Ankle Cushion Heel prosthetic foot was developed in the 1950's at the University of California at Berkley and released into the North American market in 1957 (Rose & Gamble 1996). Since that time it has become the most prescribed prosthetic foot in North America (Davies, Friz, et al. 1970). Torburn et al. (1990) used the SACH as a benchmark comparison foot stating that "...the conventional SACH foot for years has been the industry standard....". This view of the SACH was also shared by Bartkus et al. (1994) who stated "The industry benchmark (and lowest cost) prosthetic limb is comprised of a SACH foot with a steel tube pylon" (Bartkus, Colvin, et al. 1994).

Most recent comparison studies investigating the efficiency and gait patterns of new dynamic response feet such as the Flex Foot and Seattle foot have also used the SACH foot (Lehmann, Price, et al. 1993) (Torburn, Perry, et al. 1990) (Menard et al. 1992) (Casillas, Dulieu, et al. 1995) (Powers, Boyd, et al. 1996) (Snyder, Powers, et al. 1995) (Prince, Winter, et al. 1998) (Czerniecki, Gitter, et al. 1991)(Mizuno 1992). The SACH foot is also the most widely used foot in non-comparison based studies of BK gait. Many early studies measuring differences in temporal and distance parameters between amputee and non-amputee gait used the SACH foot (Skinner & Effeney 1985). For example, Robinson et al.(1977) investigated the temporal distance parameters of 19 volunteers who had been fitted with the SACH prosthetic foot. The subjects were instrumented with footswitches and accelerometers to collect stride length, walking velocity, cadence, and step-time ratio data. Results from the study showed high correlations between several time distance parameters. As well Lemaire et al. (1993) used the SACH foot when investigating the effects of age on gait patterns.

Not all comparison studies have used the SACH foot as their benchmark. Perry et al. (1975) used the single axis foot when comparing the weight acceptance mechanics of the Seattle Lite and Flex Foot designs (Perry, Boyd, et al. 1997). The single axis foot is the most widely prescribed foot in North America, second only to the SACH foot (Davies, Friz, et al. 1970). The single axis foot was more widely used in Britain and European countries during the 1980's (Goh, Solomonidis, et al. 1984).

The new low cost NPO foot design is targeted for use in 3rd world countries. At present the SACH foot is by far the most popular model of prosthetic in underdeveloped nations. Besides its relative low cost, the SACH is also easy to make from local materials

and requires little maintenance (Girling & Commings 1972) (Golding 1967) (Kijkusol 1986) (Meanley 1995) (Mensch 1986) (Pe 1988).

In 1980 a flexible keel was added to the standard SACH foot design to allow easier rollover during toe-off. The new foot was introduced in North America as the Stationary Ankle Flexible Endoskeleton foot (SAFE) (Rose & Gamble 1996). There has been an increase in the prescription of the SAFE foot over its predecessor: the standard SACH (Bateni 1996). Although the SAFE foot has a flexible keel it is still considered to be a SACH category foot (Bateni 1996). A recent study investigating impact absorption in running gait for BK amputees suggests that there may be differences between the gait patterns produced by the SACH and SAFE feet at running speeds, but no statistical significance was found (Farber & Moreinis 1995).

2.4 Factors Affecting Amputee Gait Patterns

Many factors can affect the gait patterns of BK amputees. The present study will determine if differences exist between the gait patterns produced using the two prosthetic feet. Given this purpose and the small sample size it was important to understand what other factors might account for gait pattern changes.

2.4.1 Age of Subjects

A number of studies have shown that a subject's age can affect their gait pattern. Waters (1976) examined the metabolic and gait temporal distance characteristics in ten BK amputees of various ages who walked six meters at a self-selected pace. Waters found that his elderly subjects (those over the age of 60) had statistically different gait

velocities and stride lengths when compared to his younger subjects (Waters, Perry, et al. 1976). Other studies have supported Waters' (1976) results suggesting that elderly BK amputees have lower gait velocities (Barth, Schummacher, et al. 1992) (Prince, Winter, et al. 1998) (Murray, Kory, et al. 1969). Lemaire et al. (1993) used cinematography to collect data from the contralateral legs of eight elderly BK amputees during level walking at self-selected paces. Lemaire found walking velocities and average stride length to be comparable to or above those reported in previous studies. Lemaire concluded that his inability to find differences related to age may have been due to the limitations of the prosthetic feet used and that as new prosthetic feet become more efficient, age related differences in velocity and stride length may become apparent (Lemaire, Fisher, et al. 1993). Many recent studies control the age of subjects thus preventing differences from being masked by age related gait variations.

2.4.2 Gait Speed

Walking speed has been shown to affect the temporal distance gait characteristics, kinematics, and kinetics of amputee gait. Collins et al. (1989) examined the relationship between several time distance parameters and GRF loading rates. Subjects walked at five different self-selected walking speeds while ground reaction force (GRF) data was collected using a Kiestler ™ force plate. The results revealed that altering the speed of gait has a direct affects on GRF loading rates. Collins also concluded that the relationship between cadence and the GRF loading rate at heel strike was non-linear. Robinson et. al (1977) collected gait data from 19 unilateral BK amputees and found high correlations between several of the temporal distance parameters including cadence

and step length (Hurley, McKenney, et al. 1990). Later work by Winter et al. (1983) controlled the cadence at which subjects walked. Fifteen subjects walked at high, self-selected, and low cadences while motion and force data were collected using a cinecamera and force plate. Results showed that joint moments as well as power absorption / generation varied with the cadence chosen (Winter 1983). The influences of speed on gait parameters was reported by Isakov (1996). Isakov had fourteen BK amputee volunteers walk at both a self-selected "comfortable" speed and a faster pace. From the results Isakov concluded that symmetry of gait and all temporal distance parameters are affected by the speed of gait (Isakov, Burger, et al. 1996).

Although it has not been proven experimentally, many researchers hypothesize that the Preferred Transition Speed (PTS) or self-selected pace is an energy saving mechanism in which the gait speed that optimizes energy consumption is unconsciously selected (Li, Crompton, et al. 1996) (Cavanagh & Kram 1985). Work by Hrelljac (1994) suggests that kinematic factors perceived through proprioception, such as ankle angle, may be a determinate of PTS (Hreljac 1995). Early work by Ganguli et al. (1975) compared the energy expenditure of BK amputees walking at three different gait speeds (3.0 km/hr, 4.0 km/hr, 5.0 km/hr). Ganguli concluded that since energy expenditure was minimized at 3.0 km/hr this was therefore the "optimal" speed for amputee gait (Ganguli 1975). More recent studies have suggested that amputees will "self-select" a pace at which energy expenditure is minimized (Sanders 1986). Studies have also investigated gait velocity as a measure of prosthetic performance, suggesting that increased gait velocity while wearing a prosthesis indicates improve biomechanical efficiency and confidence in the prosthetic foot. For these reasons the majority of recent prosthetic gait

studies have allowed subjects to walk at a self-selected pace rather than attempting to force a specified gait velocity.

2.4.3 Prosthetic Adjustment and Fit

Modular BK prosthetics allow for more flexibility in fit and alignment of the limb on the amputee patient and can affect the amputee's gait. The coupling components between the socket and pylon and between the pylon and foot allow adjustment of the prosthetic about three axes. These couplings can adjust the eversion/inversion, flexion/extension, and rotation of the foot with respect to the pylon and of the pylon with respect to the socket. Proper fit and alignment are essential to produce adequate gait.

In a study of the effects of the mass of the prosthesis on energy expenditure, Bateni (1996) used an alignment jig to ensure consistent alignment. The jig allowed the researchers to disassemble the prosthesis, change the components and reassemble it while maintaining the existing alignment. Similarly, when studying amputee gait patterns of dynamic elastic response feet, Torburn et al. (1990) used a vertical fabrication jig to duplicate each alignment precisely when more than the interchange of the foot-bolt and foot was required.

Visual inspection of gait is the most common method used in aligning a prosthetic limb and techniques and criteria for proper alignment vary among prosthetists (Rubin, Ficher, et al. 1986). It has been shown that changing the alignment of a prosthesis can affect gait patterns (Hannah, Morrison, et al. 1984). Not all researchers have elected to use a jig. While studying the mechanical efficiency of three different feet, Prince et al. (1998) did not use an alignment jig to align the feet on 5 subjects. The exact alignment

was not controlled, but using the same prosthetist throughout provided some consistency in alignment and fit of the prosthetic feet, suggesting that use of a more stringent control such as a jig is not necessary.

2.4.4 Accommodation

When an amputee is fitted with a new prosthetic foot they require time to become accustomed to walking with it. A subject's gait pattern will change when they first use a new foot and an accommodation period is required so the gait pattern can stabilize (English, Hubbard, et al. 1995).

English et al. (1995) studied the consistency of gait patterns in Above Knee (AK) amputees using new artificial knee joints over a four week period. They concluded that stable gait patterns were achieved after three weeks of wear and that clinical decisions on the effectiveness of new prosthetic components should not be made with an adaptation period of less than one week (English, Hubbard, et al. 1995). A review of currently published prosthetic studies shows that a range of accommodation periods have been used. The majority of experiments with comparison study designs that were reviewed allowed adaptation periods of between one day and two weeks(Hunter, Smith, et al. 1995)(Godfrey, Brett, et al. 1977)(English, Hubbard, et al. 1995)(Snyder, Powers, et al. 1995)(Allard, Trudeau, et al. 1995)(Gitter, Czerniecki, et al. 1991)(Bateni 1996). When an amputee is fitted and tested with a new foot, some researchers have allowed as much as one month for the subjects to accommodate to gait in the new prosthetic foot (Torburn, Perry, et al. 1990) (Snyder, Powers, et al. 1995). Other studies have allowed as little as

15 minutes on a new prosthetic foot to allow for gait patterns to stabilize (Prince, Winter, et al. 1998).

2.4.5 Use of Footwear

The majority of prosthetic feet designed for the North American market are intended for use inside footwear, with several notable exceptions including the Flex Foot running foot. It is possible to perform gait analysis of a prosthetic foot with a shoe or without a shoe and several studies have investigated the effect of footwear on gait patterns in BK amputees. The footwear of four of subjects was varied to determine possible effects on the GRF data from 23 male unilateral BK amputee subjects. Subjects were tested while barefoot and while wearing three different types of sport shoes. Several conclusions were drawn from these results. While walking barefoot, the GRF exhibited high frequency noise while the compliant heel of the sport shoes acted to filter the GRF. The shoe can also rigidly hold the foot ankle complex altering its response to GRF. The investigator concluded that "... the use of footwear introduces a very complex mechanism of force transmission involving: alteration of the geometry of the foot, introduction of a softer or harder phase and a combination of friction-viscous-elastic shoe-foot interface" (Seliktar & Mizrahi 1986).

The use of footwear was also investigated by Van Jaarsveld et al. (1990). Gait data were collected from five BK amputees wearing either sports shoes or leather shoes. Results showed that the magnitude of accelerations in the axial direction along the prosthetic tube at heel strike were significantly lower for subjects walking in sports shoes (Van Jaarsveld, Grootenboer, et al. 1990). These results indicate that to eliminate

differences caused by different footwear, subjects in studies evaluating prosthetic feet should wear the same footwear or none at all.

2.5 Below Knee Prosthetic Gait Analysis

The majority of prosthetic research conducted before the 1950s was largely qualitative (Hurley, McKenney, et al. 1990) and prosthetic designs were evaluated solely by subjective feedback from amputees. Although subjective feedback continues to be a valuable source of information, researchers and prosthetists require more objective data focusing on the functional performance of the feet. Currently there are three quantitative methods to analyze gait. These are:

1) Metabolic cost of gait (oxygen consumption)
2) Lower limb muscular activity during gait (EMG)
3) Evaluation of biomechanical characteristics of gait (temporal distance characteristics, kinematics, kinetics)

Some studies evaluate prosthetics using only one of the above mentioned measures while others have used a combination of all three.

The mechanical properties of a prosthetic foot can affect the energy cost of gait. Because of this many new prosthetic feet are Dynamic Elastic Response (DER) or Energy Storing Feet (ESF) that store and return energy through the mechanical deformation of the foot, thus, hypothetically, reducing the metabolic costs of gait. Several studies have measured metabolic cost to determine the validity of these claims.

Postema et al. (1997) compared the performance of two "energy storing" and two conventional prosthetic feet. The results suggested that at most a 3% metabolic cost saving was associated with the two ESF (Postema, Hermens, et al. 1997).

Mechanical efficiency of a prosthetic foot is defined as the ratio of energy loaded into the foot (usually through deformation) divided by energy out (or returned). Researchers have used devices such as the Instron to load prosthetic feet to known values and measure the return of this energy (Waters, Perry, et al. 1976) (Goh, Solomonidis, et al. 1984) (Ziolo 1999).

When it has been shown that one foot is more mechanically efficient than another, yet the total metabolic cost of walking in the foot is higher or not significantly different from other less mechanically efficient feet, researchers may use electromyography (EMG) to determine what contributes to the increased energy consumption. Rao et al. (1998) found that despite increased energy return properties from DER feet such as the Flex Foot and Seattle foot, the metabolic cost of walking in these feet was still significantly greater than able-bodied gait. The results from this EMG study of three feet (Flex-foot, Seattle, and SACH) suggested that the increased energy costs were due to increased muscle activity during the early weight acceptance stage of gait (Rao, Boyd, et al. 1998).

Considering the metabolic cost, amputee gait, regardless of foot type, is more costly than able-bodied gait. Increased energy consumption during amputee gait is due to loss of musculature and increased gait asymmetry. Less efficient asymmetric gait patterns rely on increased muscular effort from the sound limb to compensate for deficient power production from the residual limb (Waters, Perry, et al. 1976)(Huang, Jackson, et al.

1979). The gap in energy cost between amputees and able-bodied subjects, measured by O_2 consumption, increases with increased gait velocities (Hunter, Smith, et al. 1995).

Other studies have measured the metabolic cost of gait using a prosthetic foot to assess performance. Lehman et al. (1993) compared O_2 consumption during level walking of nine subjects wearing the Flex Foot, Seattle foot and standard SACH prosthetic foot. Lehman found no difference in metabolic cost between the feet (Lehmann, Price, et al. 1993). Although few studies have been able to find significant differences between prosthetic feet based on metabolic cost, the rationale behind collection of this data is sound..

By far the most popular means of assessing prosthetic gait performance is through evaluations of biomechanical characteristics. Biomechanical gait characteristics can be grouped into three major categories:

1) Time Distance Parameters
2) Motion Analysis
3) Ground Reaction Forces (Force Plate)

2.5.1 Time Distance Parameters

Time distance parameters are also referred to in literature as temporal distance factors, stride characteristics, or spaciotemporal characteristics. These variables are the most common variables used to assess performance in gait studies because they require minimal equipment (Skinner & Effeney 1985). Many of these variables could be

collected using a simple footswitch and a stop watch. The five most commonly reported time distance parameters are:

1) Gait velocity
2) Cadence
3) Step length
4) Stance ratio
5) Stance time (or double support time)

Several time-distance parameters are highly correlated. For example, gait velocity and cadence are highly correlated (0.88), as have gait velocity and step length (0.83) (Robinson, Smidt, et al. 1977). On the other hand, there is only a low correlation between cadence and step length (Hurley, McKenney, et al. 1990).

Many gait studies have shown differences in time distance parameters between amputee and able-bodied subjects. For example, the gait velocity of AK amputees can be up to 38% slower than that of able-bodied subjects (James & Oberg 1973) (Murray, Mollinger, et al. 1983) (Skinner & Effeney 1985). Waters et al. (1976) calculated time distance characteristics for AK and BK amputees and found that free walking velocity, cadence and stride length for both groups were lower than that of able-bodies subjects (Waters, Perry, et al. 1976). Other studies have confirmed these results (Ganguli 1975) (Huang, Jackson, et al. 1979). Step length and swing to stance ratio have been found to be significantly different between BK and able-bodied subjects as well as between the amputee's sound and residual limbs (Hurley, McKenney, et al. 1990). Amputees take longer steps with their residual limb as compared to their sound limb (68cm and 63 cm

respectively). The mean percentage of the gait cycle time spent in the stance phase for the residual limb was 61%. This value was lower than the mean time spent on the sound limb (65%). Murray et al. (1966) performed gait trials on 10 amputees and confirmed these findings, obtaining lower velocities, longer gait cycles and slightly shorter stride lengths when compared to controls (Murray, Kory, et al. 1966). These findings demonstrated that asymmetries exist between legs in amputee gait (Hurley, McKenney, et al. 1990).

Gait velocity is a good determinate of prosthetic performance because it has been shown to correlate with joint impairment and other acute problems in the lower extremities (Skinner & Effeney 1985). Increased GRF at the joints of the sound limb may be due to the lack of calf musculature and the loss of an articulating ankle joint. Ankle dorsiflexion allows the movement of the center of gravity to be minimized in the vertical plane, while contraction of the calf muscles controls deceleration of the body during fore-aft motion. Loss of these functions would cause an increased vertical and fore aft GRF seen in the sound limb that may cause increased joint degradation (Simon 1985).

It has been suggested that a decrease in gait velocity is a strategy to reduce the GRF acting on the sound limb to an acceptable level (Hurley, McKenney, et al. 1990) (Lewallen, Dyck, et al. 1986). A second hypothesis is that a decreased gait velocity allows the amputee to minimize energy expenditure when walking on inadequate or inefficient prosthetic foots. Breakey (1976) supports this hypothesis, suggesting that increased gait speeds are evidence of a more efficient prosthetic foot, allowing a more symmetrical gait (Breakey 1976).

Stride length and swing-stance ratio are variables that allow the researcher to determine the amount of asymmetry that exists between legs during gait. Decreased symmetry will increase the excursion of the center of mass in both the vertical and fore aft directions when walking regardless of foot type. The absence of shank muscles decreases the subject's control over this excursion thereby increasing the energy cost of amputee gait (Skinner & Effeney 1985). Stride length has been shown to differ between prosthetic feet suggesting changes in symmetry with different feet. Powers (1997) was able to show a significant difference in stride length of the Flex Foot when compared to four other components (SACH, Carbon copy II, Seattle, Quantum), suggesting improved gait performance. Both swing-stance ratio and stride length are valuable variables for determining performance differences between prosthetic feet.

Another equally important time distance variable is gait velocity. Perry et al. (1997) showed significant differences between the gait velocities of the single axis, Seattle and Flex Foot designs during a weight acceptance study. The Flex Foot design allowed amputees to walk at higher gait velocities. (Perry, Boyd, et al. 1997).

During a study examining the weight acceptance phase, the Flex Foot allowed the amputee to walk faster than the single axis or Seattle foot (Perry, Boyd, et al. 1997). However, one must be cautious when drawing conclusions based on velocity alone. Studies have shown that increased strength of the quadricep muscles can also increase gait velocity. Below knee amputees who completed a muscle strengthening program increased gait speeds by up to 13%. This suggests that some variation in gait velocity is due to changes in quadriceps strength and may not be due to foot performance alone. (Kegel, Burgess, et al. 1981).

2.5.2 Motion Analysis

A more complete characterization of gait patterns can be obtained by examining the motion of the lower limb segments (Skinner & Effeney 1985). To track the position of segments in space during gait researchers place markers on anatomical landmarks. At least two markers are required to define a body segment (e.g. markers placed at the lateral tibial epicondyle and lateral maleolus define the shank). The motion or the relative angle of a joint can be determined when two adjacent segments are defined, for example the knee joint is defined as the angle created between the thigh segment, and shank segment. To acquire three dimensional data a third marker is required (Allard, Stokes, et al. 1995).

The location of a segment in space during three dimensional gait analysis can be described with the use of two coordinate systems. A global (or absolute) coordinate system is a system that is independent of the subject and normally fixed in relation to the lab area. Fixed body (or relative) coordinate systems are affixed to the subject.

Researchers have used a number of different devices to track the position of a marker in space. Common systems of motion analysis include electromagnetic, acoustical, cinematography, and optical-electric systems (Allard, Stokes, et al. 1995).

Motion analysis has been found to be a sensitive technique for discerning differences in the hip, knee, and ankle angles of amputees over able-bodies subjects (Wirta & Golbranson 1980). It is known that below knee amputees have reduced knee flexion angles during the stance phase of gait (Skinner & Effeney 1985). Culham et al. (1984) used electrogoniometers to provide continuous records of knee flexion and extension angles bilaterally during the gait of ten BK amputees wearing either the SACH or single

axis prosthetic foot. The mean peak knee angles at heel strike for the SACH and Single axis during stance were 17.68 +/- 4.60^0 and 16.34 +/- 7.00^0. Culham also found differences between the feet in the peak knee flexion angle during swing (46.37 +/- 9.60^0, 41.34 +/- 7.44^0), and the timing of peaks within the gait cycle. Culham attributed these differences in kinematics to the mechanical differences of the feet since the fixed ankle joint of the SACH requires increased knee flexion to allow for toe clearance during swing (Culham, Peat, et al. 1986).

Comparisons of knee angles between residual and sound limbs can quantify the asymmetry of prosthetic gait. Greater differences in knee angle and knee angle timing increase the asymmetry of gait and are therefore indicative of a less efficient foot (Isakov, Burger, et al. 1996).

Motion analysis can also be used to determine where joint angle differences may create performance differences. Powers et al. (1994) studied the gait of ten BK amputees wearing five different prosthetic feet (SACH, Flex Foot, Carbon copy II, Seattle, and Quantum). Powers was able to show a significantly larger ankle dorsiflexion angle for the Flex Foot in terminal stance compared to the other four feet. Powers suggests that this increased ankle angle is responsible for a time delay in the maximum posterior shear force seen in the Flex Foot and may account for the smoothness in its gait pattern (Powers, Boyd, et al. 1996). Other studies have demonstrated that differences in knee flexion angles when wearing different prostheses are caused by the prosthetic foot's ability to act as a primary shock absorbing mechanism during loading of the joint. When a greater proportion of the loading is absorbed by mechanical deformation of the foot, less active knee flexion is required (Snyder, Powers, et al. 1995). Increases in knee

flexion angles during stance have also been linked to increased activity in the quadriceps muscles. The quadriceps have been shown to contract eccentrically to restrain the knee joint during flexion and prevent collapse of the joint (Perry, Boyd, et al. 1997) (Winter 1980).

Motion analysis can be used to quantify the angular velocities of body segments as well as joint angles (Allard, Stokes, et al. 1995). Prosthetic foot characteristics determined through motion analysis such as the level of asymmetry, mechanics of a foot ankle assemble, and load absorption characteristics of the joints during gait give the researcher a means of ranking feet based on performance.

2.5.3 Ground Reaction Forces

The patterns, magnitude, and timing of GRF events can be measured using a force plate. A force plate is an instrument which provides readings of forces and moments applied to its top surface while the foot of the subject is in contact with the plate (Skinner & Effeney 1985). Vertical reaction forces for BK amputees have been shown to be lower in magnitude with a smaller trough (minimum force peak following heel strike) than the GRF patterns of able-bodied subjects (Rose & Gamble 1996). Other studies have confirmed these findings showing that vertical GRF troughs tend to be lower in magnitude than able-bodied values regardless of prosthetic foot used (Powers, Boyd, et al. 1996)(Snyder, Powers, et al. 1995)(Mizuno 1992). The asymmetric nature of gait in BK amputees is also seen in the GRF patterns and show significant variations between the residual and sound legs (Suzuki 1972).

Mizuno (1992) compared the GRFs of twelve BK amputees using nine test prosthetics (Greissinger, Multiflex, Otto Bock Dynamic, SACH, SAFE, Carbon Copy II, Quantum, Seattle, Seattle Lite). Subjects were asked to walk at a comfortable (self-selected) pace across a walkway that had two force plates mounted flush to the floor. Ground reaction force data from the residual and sound legs was normalized to body weight (force/bw), averaged across both feet and compared for differences. Differences were found between feet in the depth of the GRF vertical trough and the ability of the feet to decelerate / accelerate the subject in the fore-aft plane. Average values for the vertical troughs were 28.87 +/- 12.398 (%BW), and 46.23 +/- 4.600 (%BW) for the residual and sound limb respectively. Efficiency of deceleration / acceleration values for residual and sound limbs were reported as 29.31 +/-5.56 (%BW) and 45.85 +/- 3.93 (%BW) respectively (Mizuno 1992).

Compensatory mechanisms employed by the sound limb can mask GRF abnormalities of the residual limb. Studies have shown that due to deficiencies in the ability of the residual limb to absorb impact, the sound limb is susceptible to increased vertical forces during loading (Powers, Boyd, et al. 1996) (Engsberg, Lee, et al. 1993). When sound limb GRFs were compared during gait in four prosthetic feet Snyder et. al (1995) demonstrated that the sound limb accepted 11% more body weight during loading (Snyder, Powers, et al. 1995). GRF data from both the residual and sound limbs allows researchers to determine what, if any, effects a prosthetic foot may have on the gait patterns of either limb.

Although GRF patterns are an undeniably excellent tool for discerning differences between prosthetic feet, some researchers have cautioned against adopting conclusions

based on this data. This is because variation in GRF data between prosthetic feet, between BK subjects and between trials for amputees is higher than that in able-bodied subjects (Seliktar & Mizrahi 1986). These researchers have pointed out that factors such as compensatory movements of the trunk or upper limbs can also affect GRF characteristics (Seliktar & Mizrahi 1986) (Menard 1988).

2.5.4 Joint Moments

The three properties of a body segment required to calculate moment and force are mass, center of mass location, and mass moment of inertia. In experiments using able-bodied subjects these values are often calculated using regression equations derived from studies using cadavers. The properties of artificial limbs do not conform to those of normal limbs, and must be either estimated through modeling of the prosthesis and residual limb, or measured directly.

The peak moments observed at the joints of the residual limb (hip, knee, and ankle) of BK amputees during level gait have been shown to be lower than those of able-bodied individuals (Smidt 1990). Studies have also suggested that joint moments during level gait are more variable within the BK population compared with able-bodied populations (Smidt 1990). Moments acting at the knee joint of the residual limb tend to be lower than those of able-bodied subjects and were close to zero through a large proportion of gait cycle. Winter et al. (1988) attributed the change in knee moment pattern partly to an increased ankle dorsiflexion moment during early stance and partly to the ankle joint of the SACH foot which is solid and does not allow normal dorsiflexion or forward rotation of the foot towards the ground. As the body progresses forward over the foot an internal

ankle dorsiflexion moment is created that counters the normal moment at the knee joint (Prince, Winter, et al. 1998). Results from other studies have supported findings of Winter (1988), showing that the SACH exhibits decreased dorsiflexion angles during early stance (Lewallen, Dyck, et al. 1986) (Torburn, Perry, et al. 1990). Lewallen et al. (1985) showed that BK amputees also experience smaller plantarflexion angles and moments during late stance phase due to a prosthetic foot's relative inflexibility in the forefoot and a lack of plantarflexion muscles (Lewallen, Dyck, et al. 1986). The sagittal plane hip, knee and ankle moments create a stable support moment and changes in the moment at one joint must be counterbalanced by alterations in the moments at the other joints to maintain proper support (Winter 1980). This may account for the moment pattern differences between amputees and able-bodied individuals.

Studies have also found asymmetries in the magnitude and pattern of the joint moments of the sound and residual limbs (Winter & Sienko 1988). Moments at the knee and ankle are generally higher on the sound limb (Powers, Boyd, et al. 1996). Robinson et al. (1977) reported that step length of the BK amputees was longer on the residual side and was accomplished in a shorter time. This indicates an increased acceleration of the residual limb that must be counterbalanced by the actions of the sound limb, and may account for increases in knee moments on that side (Robinson, Smidt, et al. 1977)(Powers, Boyd, et al. 1996). This may account for the differing knee extension and ankle dorsiflexion moments (Lewallen, Dyck, et al. 1986). Plantarflexion moments at the ankle during late stance create forward propulsive impulse. Decreased ankle moments at the residual limb must be balanced by increased moments at the sound limb leading to increased asymmetry of gait (Lewallen, Dyck, et al. 1986).

2.5.5 Power Absorption / Generation

Negative power indicates an eccentric contraction removing energy from the system while positive power indicates a concentric contraction adding energy to the system. The power bursts created at the joints during gait are:

A1 – absorption by plantarflexors as the leg rotates forward over flat foot.

A2- generation by plantarflexors (push-off) as the foot plantarflexes prior to toe-off.

K1- absorption by knee extensors as the knee flexes during weight acceptance.

K2- generation by knee extensors as the knee extends during mid stance to raise the center of gravity of the body.

K3- absorption by knee extensors during push-off as the knee flexes prior to and after toe-off

K4- absorption by knee flexors at end of swing to take out energy of swinging leg and foot.

H1-brief generation by hip extensors at weight acceptance as the hip extends (as knee flexes).

H2- absorption by hip flexors to decelerate backward rotation thigh.

H3- generation by hip flexors as hip flexes before toe-off and in early swing to pull the lower limb upwards and forward; this action is now referred to as pull-off (as opposed to push-off by the plantarflexors).

Winter et al. (1991) defines the K1, K3, K4, and A1 phases as the major sources of power absorption while H1, H3, K2 and A2, are the major sources of the power generation (Winter 1991). Later work by Winter et al. (1988) suggests that the major phases of power generation in amputee gait differ from able-bodied gait (Winter & Sienko 1988). Other studies have shown that the hip musculature has increased involvement in power absorption and generation in the residual limb due to lack of the muscles of the shank. (Torburn, Perry, et al. 1990) (Czerniecki, Gitter, et al. 1991). Asymmetries in joint power production between legs have been shown to be a compensatory mechanism of BK amputees. Increased power absorption and generation of the sound limb allows compensation for deficits in muscle ability in the residual limb (Gitter, Czerniecki, et al. 1991).

Observations of the amputees' ability to produce power with the residual limb, and the magnitude of the asymmetry between limbs are valuable means of assessing prosthetic foot performance.

3.0 Methods

3.1 Pilot Testing

A protocol for a comparison based study to assess the gait of BK amputees using a new prosthesis was developed from a review of current literature. Pilot testing ensured that this protocol was sound and that no modifications of the NPO prosthesis were required (Appendix A).

3.2 Subject Recruitment

Subject recruitment was performed through the prosthetics department at St. Mary's on the Lake Hospital. Head prosthetist Robert Merritt was provided with the subject inclusion criteria developed for the study (see 3.2.1 Subject Inclusion Criteria). Mr. Merritt reviewed his current patient files to find subjects matching the criteria. Subjects were invited to meet with researchers at St. Mary's hospital during their normally scheduled prosthetics appointments to discuss their participation in the study.

3.2.1 Subject Inclusion Criteria

Subject inclusion criteria were developed to minimize gait variations introduced due to subject differences and pathologies of gait. Inclusion criteria included the following:

1) Subjects were young adults – 18 years old to 55 years old.
2) Subjects were active and could walk without support (i.e. cane).
3) The time since amputation was to be a minimum of two years.

4) Subjects had no stump abnormalities – the subject's residual limb had a stable volume, with no skin sores and no significant bone deterioration.
5) Subjects were to be "good walkers" – The prosthetist in charge of their care determined that each subject exhibited a normal BK gait pattern.

Five of the seven unilateral BK amputees who met the inclusion criteria volunteered to participate in the study. These subjects read and signed a consent form. It outlined the purpose of the study, provided an overview of the testing procedure and the subject's rights and responsibilities (Appendix B). Each subject was assigned a code to ensure that confidentiality was maintained.

3.3 Pre-Trial Accommodation

Subjects were required to attend two testing sessions at the gait laboratory located in the Kingston General Hospital (KGH). At each session a different foot was tested. Prior to each testing session the subject was provided with either a SAFE or a NPO foot and allowed walk with the prosthesis for a period of no less than two days. During session one, the subject's gait was assessed while wearing the SAFE prosthesis while the NPO prosthesis was assessed during session two. The required prosthetic foot sizes were obtained from patient records at St. Mary's and provided to the researchers by Mr. Merritt. The prosthetics department at St. Mary's hospital supplied SAFE prosthetic feet for each subject. Queen's Clinical Mechanics Group provided the NPO prosthetic foot.

The NPO feet were prototypes, not production models. An engineering student with the Queen's Clinical Mechanics Group (CMG), Tara Ziolo, developed a finite element

model of the NPO prosthesis for mechanical testing (Ziolo 1999). If the foot length was input into the model the remaining dimensions of the foot were automatically scaled to produce a computer model of the foot. In this way, the models for each foot size were supplied to the Hia Precision Cutting Company (Trenton, ON) who cut the NPO prosthetic feet from blocks of nylon 66 using a water jet cutter.

The prosthetic feet (both NPO, and SAFE) were fitted and aligned for each subject by Mr. Robert Merritt (CP&O). The subjects then wore the foot for at least two days to allow for some accommodation.

3.4 Equipment

The Queen's Gait Analysis in Three Dimensions (QGAIT) system developed at Queen's university was used to collect and process gait data from subjects. The QGAIT system software integrates kinematic, kinetic and anthropometric information to calculate angles, forces and moments at the joints of the lower limb (Costigan, Wyss, et al. 1992)(Li, Wyss, et al. 1993).

An optoelectric motion tracking system (Optotrak) from Northern Digital was used to acquire three dimensional kinematic data. The Optotrak system consists of a sensor camera, supporting electronics, a series of Infra-Red Emitting Diodes (IRED) and a collection software package from Northern Digital (Northern Digital Inc. 1992). The Optotrak camera has three infra-red sensitive lenses and is mounted vertically to the wall of the gait laboratory approximately four meters from a floor mounted force plate (figure 3.1).

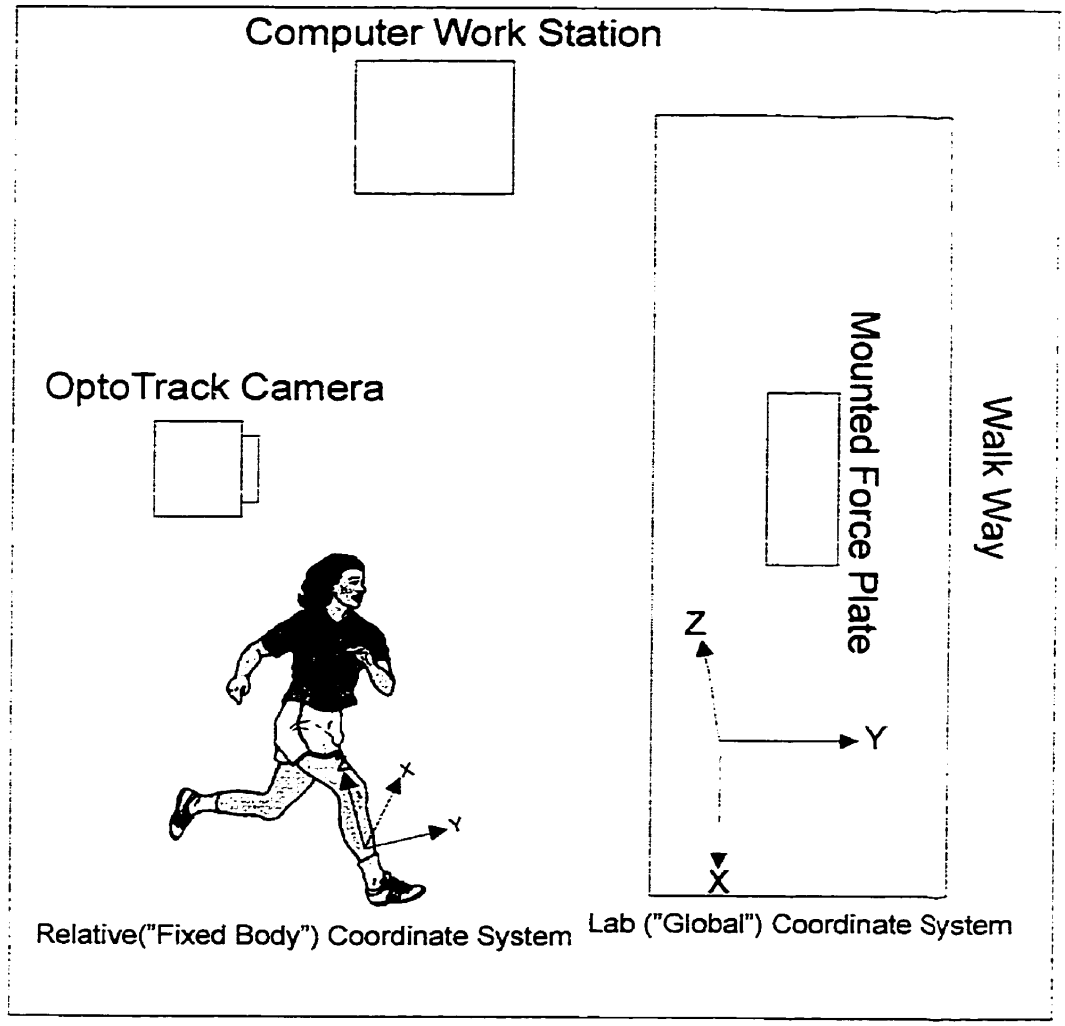


Figure 3.1: Gait lab setup

An AMTI (Advanced Medical Technology Inc.) force plate mounted flush with the floor in the center of the 6 meter walkway measured the ground reaction force. The AMTI LG-4060 force plate is a strain gauge type force plate containing four load cells in a pillar configuration (one load cell under each of the four corners). GRF data were collected through a 16 channel 12 bit analog-to-digital conversion board supplied by

Northern Digital (Waterloo, Ont.). The Optotrak system unit allowed the simultaneous collection of the kinematic and force plate data.

For the Optotrak and force plate to share a common fixed coordinate system a calibration must be performed before the collection of gait data. A cubic steel frame containing 24 IREDs was used for calibration. The frame was aligned with edges of the force plate so that the force plate and motion coordinate systems were aligned.

A manual button was used to determine heel strike during gait. The button was depressed each time the subject's heel made contact with floor generating a voltage spike on one channel attached to the analog to digital board. This spike was collected simultaneously with the Optotrak and force plate data and was used to determine the timing of the gait cycle.

3.5 Subject Preparation

Upon arrival to the gait lab the testing procedure was reviewed with each subject. Initial anthropometric information was recorded for each subject that included height and weight. The subjects were provided with athletic shorts, and asked to remove their shoes and socks from both feet.

3.5.1 Subject Instrumentation

A series of 12 IRED markers were affixed to the subject's residual limb (table 3.1). The first nine of these markers corresponded to anatomical landmarks of the lower extremities. Probes projecting away from the body were attached to the subject's thigh

and shank. Each probe had an IRED marker at its end and was a third marker on each of the thigh and shank. These three markers were required to determine a segment's spatial orientation (figure 3.2).

Marker locations for the residual limb were estimated from the sound limb if anatomy was not present (e.g. the malleolus bone marker location on residual limb was estimated from sound limb's location). A series of three IRED markers were affixed in a triangular formation on the subjects back at the level of L5 and allowed the calculation of hip joint angles.

Table 3.1: Location of landmarks

Marker No#	Marker Location
1	Greater trochanter
2	Thigh probe (raised)
3	Lateral inferior epicondyle of the femur
4	Lateral superior epicondyle of the tibia
5	Shank probe (raised)
6	Lateral malleolus
7	Lateral calcaneus
8	5^{th} metatarsal
9	5^{th} toe
10	Back probe 1 (L5 level)
11	Back probe 2 (L5 level)
12	Back probe 3 (L5 level)

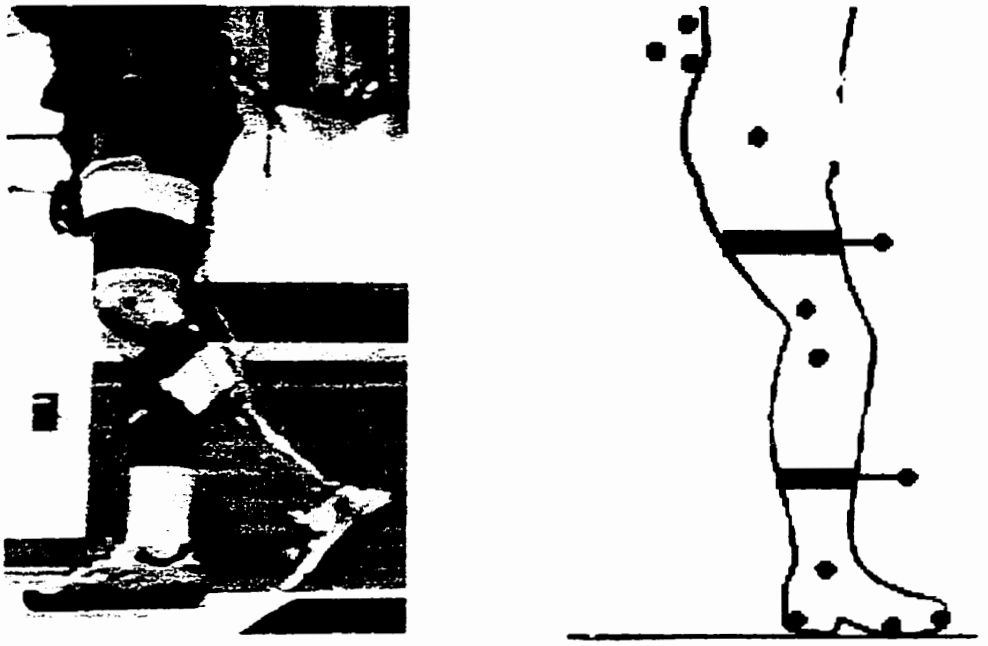


Figure 3.2: Location of IRED markers

Prior to the gait trials the subject stood on the force plate with their foot aligned at a 90 degree angle (perpendicular) to the Optotrak camera and a static position was collected. This static collection creates a reference position that is used to correct knee and hip angles. The reference position also allows the determination of a limb coordinate system so that surface markers can be moved into the segments. During this time each IRED marker was checked to ensure that they were functioning and visible to the Optotrak camera.

3.6 Gait Trails

Subjects walked at a self-selected pace along the walkway. Gait data were collected as the subject walked across the force plate in the middle portion of the walkway. This ensured that the subject was neither accelerating nor decelerating but had reached a constant walking velocity.

Before collection began, the subject performed several trial walks to become familiar with the instrumentation. While walking, the subjects held their arm across their chest - purse carry style- to ensure that the greater trochanter IRED was not obscured. Subjects were instructed to look directly forward to prevent targeting of the force plate. The subject's starting position was adjusted so that proper foot contact was made on the force plate. Proper foot contact was achieved when the subject's affected foot and only the affected foot made full contact with the plate. Once a suitable starting position was located it was marked with tape to ensure a consistent start point. Each time the subject's heel made contact with the floor a button was pressed that sent a signal to the analog-to-digital converter. This signal indicated the heel strike event and was used to define the step cycle.

Data from the Optotrak and force plate were collected simultaneously at a sampling rate of 100 samples per second (100 Hz) for a period of four seconds. The force plate amplifier was set to a gain of 1, with a frequency filter of 1040 Hz. Collection of data was started on the heel strike preceding force plate contact. A collection period of four seconds was set to guarantee that at least one full gait cycle would be captured.

The subjects performed seven walking trials. Trials were visually inspected so that: 1) no more than 3 consecutive data points were absent for any one IRED, 2) missing data did not occur at curve inflexion points, and 3) the force plate profiles were consistent. If less than five of the seven trials were acceptable then additional trials were collected.

Once five satisfactory trials were collected, anthropometric data were measured and correction vectors for the IREDs were estimated. These values were recorded in the anthropometric and correction vector data sheet (Appendix C). The markers were removed from the residual limb and transferred to the sound side. The collection protocol was repeated for the sound limb and was identical to that described for the residual limb.

To calculate joint forces and moments of the sound limb the QGAIT system software estimates segmental weights, centers of mass, and mass moments of inertia for segments base on a series of regression equation derived from cadaver studies (Li, Wyss, et al. 1993). Measures of the lengths from floor to the greater trochanter IRED, and floor to tibial plateau IRED, as well as the circumference of the upper thigh, and calf were collected and entered in the anthropometric and correction vector data collection sheet (Appendix C).

3.7 Correction Vectors

Markers on the skin surface do not represent the limb endpoints and as such needed to be corrected so that three-dimensional forces and moments can be calculated. Estimations of the distance of the marker to the location of the joint center (hip, knee, and ankle) were found using calipers. True hip center in both the Posterior-Anterior and Distal-Proximal axes was estimated by half the distance in those planes between the

marker and the anterior superior ilio-sacral joint. The marker on the lateral epicondyle of the femur was move medially half the distance measured between the medial and lateral epicondyles of the femur. The marker was also corrected distally to meet the joint line of the knee. The fibular head marker was corrected medially half the distance measured between the fibular head and the medial epicondyle of the tibia and proximally to meet the joint line of the knee.

3.8 Subject Questionnaire

Upon completion of the gait trials each subject filled out a questionnaire regarding the prosthesis that they were wearing. The questions determined the subject's perception of the prosthesis in terms of comfort, stability, easy of use, and efficiency. Each question was rated using a ten point Likert type scale. The questionnaire had space at the end to comment on their perceptions of the foot. During the second testing session the subjects were asked to indicate which of the prosthetic feet they preferred (Appendix D).

3.9 Residual Limb Properties

The properties of the prosthesis (socket, pylon, and foot) and stump differ from normal (weight, density, center of mass etc.) so that regression equations based on normal limbs can not be used to calculate segmental weights, centers of mass location and mass moments of inertia for the prosthesis and stump. These values had to be either measured or estimated and entered into the QGAIT program.

3.9.1 Mass Fraction

The leg prosthesis was removed from the stump and weighed using a Shimpo force gauge (Shimpo American Corporation, Lincoln wood, Il). A volumetric displacement technique was used to determine the mass of the stump. A small hole was drilled near the top a large cylindrical plastic bucket creating a "spill bucket" (figure 3.3). The spill bucket was filled with water above the level of the hole and allowed to drain until the water level was immediately below the spill hole. The knee joint line was estimated for each subject by palpating the stump while the subjects flexed and extended their knee. The joint line was drawn on the subject's skin. The subject then submerged their stump in the spill bucket up to the joint line. The water displaced by the stump poured out of the spill hole and was collected in a flask. The volume obtained for the stump was multiplied by 1.09 g/LM to convert the volume to a mass (Gitter, Czerniecki, et al. 1991).

The mass of the stump was added to the mass of the leg prosthesis to give a combine mass of the both. This value was entered into the QGAIT program as the mass fraction of the shank / foot complex.

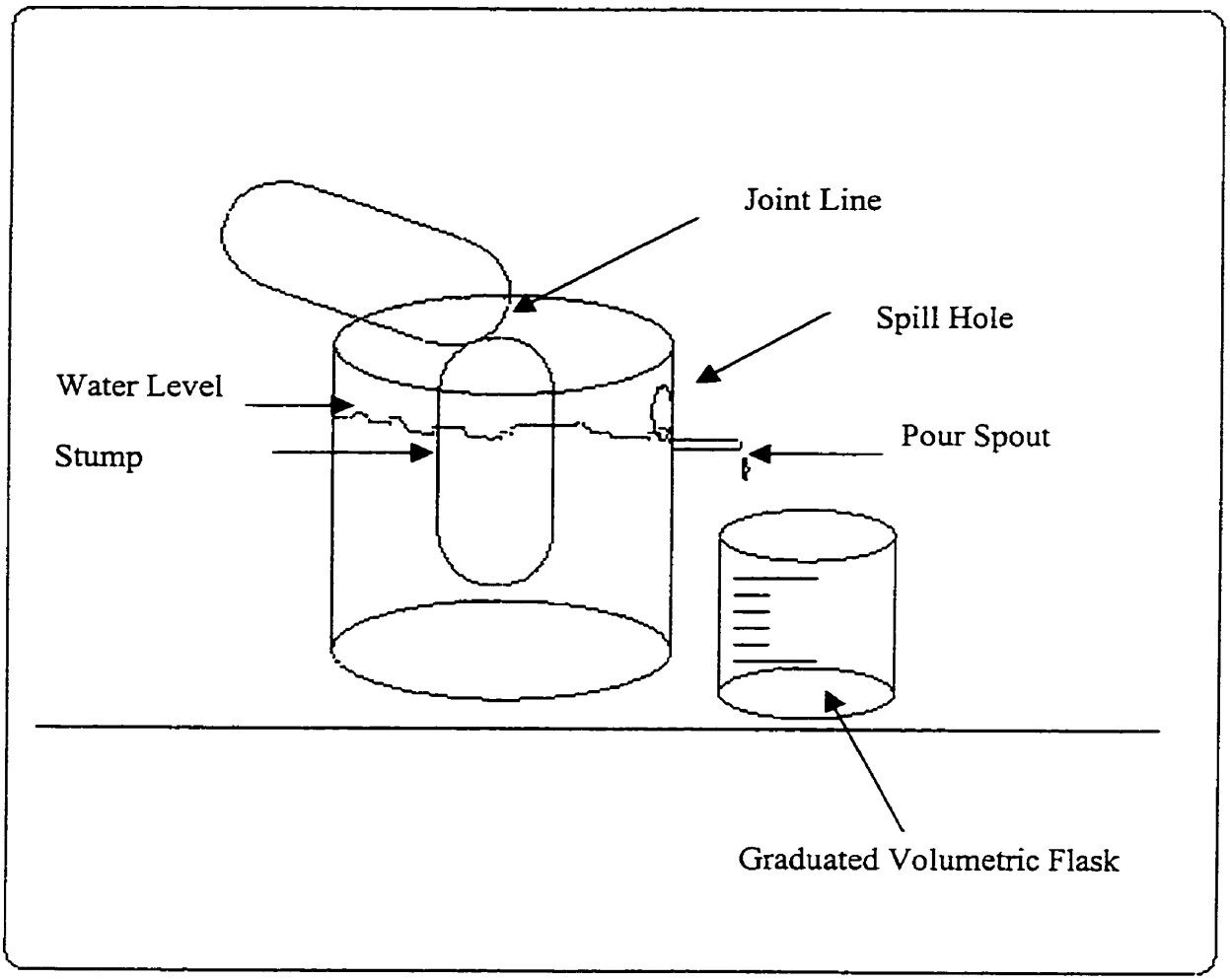


Figure 3.3: Volumetric measuring device

3.9.2 Center of Mass

To determine the center of mass location for the prosthesis a knife edge balance technique was used. The prosthesis was assumed to be symmetrical about both the medio/lateral, and posterior/anterior axes. The prosthesis was placed on the knife edge triangle and its position adjusted until a balance point was located along its distal/proximal axis (figure 3.4). A measurement from the proximal end of the prosthesis (the socket) to the balance point was taken using a tape measure.

To determine the center of mass of the stump, two measure were taken: 1) circumference of the stump at the joint line, and 2) total length of the stump from joint line to stump end. The stump was then modeled as a three dimensional parabola to determine its center of mass.

The parallel axis theory was used to find a combined center of mass for the stump and the leg prosthesis.

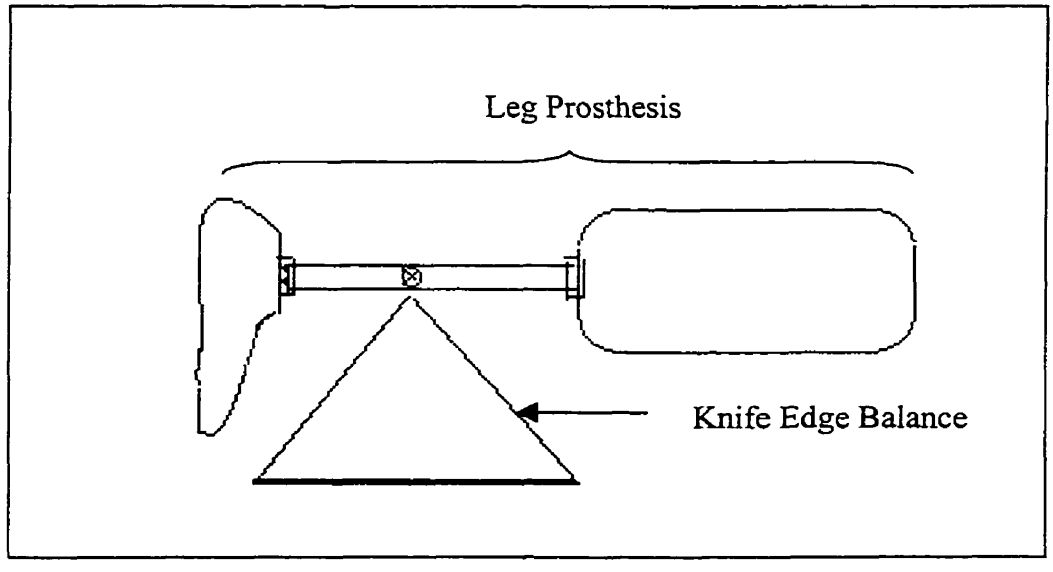


Figure 3.4: Knife edge balance setup

3.9.3 Mass Moment of Inertia

Using a near frictionless pendulum the mass moment of inertia of the leg prosthesis was estimated. The leg prosthesis was attached to the pendulum and, from a fixed height, let go to swing freely. The time for a complete swing was measured and used to calculate mass moment of inertia (see Appendix E for calculations).

The stump was modeled as a truncated cone for one subject. From this model the estimated contribution of the stump to the mass moment of the inertia of the stump, prosthesis complex was calculated. Because the contribution of the stump was much smaller than the estimated error for the prosthesis mass moment of inertia its contribution was assumed to be zero.

3.9.4 Estimation of Error

Errors in estimating the segment mass, the location of the center of mass and the mass moment of inertia on the calculated joint forces and moments were investigated. Values were chosen to represent reasonable amounts of error that could be made during these estimates. The data from a single subject was processed using five different values for each of mass fraction center of mass, and mass moment of inertia. The values used were the original value and the original value plus and minus one and two times the estimated error. The RMS of the difference between the original and error conditions were computed for the entire gait cycle for the joint forces and moments. The RMS difference was expressed as the percent difference of the range of the original curve.

RMS (force) /RMS (max force – min force) x 100% 3.1

RMS (moment) / (max moment – min moment) x 100% 3.2

3.10 Gait Data Processing

Raw gait data was processed using the QGAIT software. The QGAIT program is a batch file that calls a series of sub programs to process data. Sub programs within the QGAIT program perform the following processing tasks:

Perform linear interpolation of the data to fill in any missing data points.

Filter the data with a fourth order, zero lag butterworth filter.

Use the correction vectors to move the markers into the joint centers.

Pick start and end points of the gait cycle.

Normalize the gait cycle to 101 point (0-100%)

Calculation of forces, moments, angles, and powers

Temporal distance parameters were calculated using the TIMEDIST program (part of the QGAIT collection of programs). The TIMEDIST program uses kinematic data acquired from the Optotrak to calculate: step length, gait time, gait velocity, cadence, stance ratio, and stance time. (Costigan, Wyss, et al. 1992)

3.11 Data Analysis

Variables were separated into four categories: time distance parameters, gait curve patterns, curve parameters, subjective evaluations. A list of all variables tested from the time distance parameters, and curve parameters categories are presented in table 3.2.

Table 3.2: List of variables

TIME DISTANCE PARAMETER
Step Length
Gait Time
Gait Velocity
Cadence
Stance Ratio
Stance Time
CURVE PARAMETERS
Maximum Moment AP axis (0-40% of Gait Cycle)
Maximum Moment AP axis (40-100% of Gait Cycle)
Location of Max Moment AP axis (0-40% of Gait Cycle)
Location of Max Moment AP axis (40-100% of Gait Cycle)
Maximum Moment LM axis (0-100% of Gait Cycle)
Location of Maximum Moment LM axis (0-100% of Gait Cycle)
Range of Moments LM axis (0-100% of Gait Cycle)
Maximum Moment DP axis (0-100% of Gait Cycle)
Minimum Force AP axis (0-20% of Gait Cycle)
Maximum Force AP axis (0-100% of Gait Cycle)
Minimum Force LM axis (0-100% of Gait Cycle)
Minimum Force DP axis (0-100% of Gait Cycle)
Range of Knee angles AP axis (0-100% of Gait Cycle)
Maximum Knee angle LM axis (0-50% of Gait Cycle)
Maximum Knee angle LM axis (50-100% of Gait Cycle)
Range of Knee angles DP axis (0-100% of Gait Cycle)
Maximum Knee Power AP axis (0-20% of Gait Cycle)
Maximum Knee Power AP axis (20-100% of Gait Cycle)
Maximum Knee Power LM axis (0-100% of Gait Cycle)
Minimum Knee Power LM axis (0-100% of Gait Cycle)
Maximum Knee Power DP axis (0-100% of Gait Cycle)
Minimum Knee Power DP axis (0-100% of Gait Cycle)
Positive Force Impulse AP axis (0-100% of Gait Cycle)
Negative Force Impulse AP axis (0-100% of Gait Cycle)
Location of Force Impulse Transfer (20-50% of Gait Cycle)
Maximum Ground Reaction Force LM axis (0-30% of Gait Cycle)
Maximum Ground Reaction Force LM axis (30-60% of Gait Cycle)
Maximum Ground Reaction Force DP axis (0-30% of Gait Cycle)
Maximum Ground Reaction Force DP axis (30-60% of Gait Cycle)
Minimum Ground Reaction Force DP axis (20-50% of Gait Cycle)
Slope of Ground Reaction Force DP axis (0-11% of Gait Cycle)
Minimum Ankle Angle LM axis (0-20% of Gait Cycle)
Maximum Ankle Angle LM axis (10-60% of Gait Cycle)
Range of Ankle Angles LM axis (0-100% of Gait Cycle)

Averages and standard deviations were calculated using Microsoft Excel 2000 for all time distance and curve parameters. The five trials for each subject were average to produce a subject average. The subject averages were then averaged to provide a ensemble average.

A two way analysis of variance (ANOVA) with repeated measures was used to determine which variables from the time distance and curve parameter categories were different between feet at an alpha of 0.05. The SPSS statistical analysis software package (Ver. 8.0) was used to perform these tests. A test for observed power was also performed for each variable. The two factors tested in the ANOVA were prosthetic and trial.

4.0 Results

4.1 Subject Information

The five subjects ranged in age from 28 to 45 years. Anthropometric measurements revealed that the subjects ranged in height from 167.5 cm to 187 cm, and in weight from 72.2 kg, to 97.7 kg. One subject's amputation was as a result of disease while the remaining four amputations were performed after traumatic injury to the limb (Table 4.1).

Table 4.1: Subject Information

Subject code	Age (years)	Weight (kg)	Height (cm)	Cause of amputation
CL	45	97.7	169.5	Trauma
DK	46	96.8	187	Trauma
MA	36	79.1	167.5	Trauma
MM	34	72.7	173.5	Trauma
SG	28	90	172	Cancer

4.2 Time Distance Parameters

The stance ratio from the residual limb was the only time distance variable found to be significantly different between the NPO and SAFE prosthetic feet. The stance ratios for the NPO and SAFE feet on the residual limb were found to be 0.613 (± 0.025) and 0.647 (±0.021) respectively. Differences could not be detected for the remaining five variables. Bar graphs of for each of the six time distance parameters can be seen in Appendix F. Values average by prosthetic with their associated standard deviations can be see in table 4.2.

Table 4.2: Average time distance parameters

	Residual		Sound	
	NPO	SAFE	NPO	SAFE
Step Length (m)	1.31 (0.08)	1.24 (0.10)	1.31 (0.09)	1.28 (0.06)
Gait Time (s)	1.20 (0.06)	1.17 (0.05)	1.20 (0.05)	1.19 (0.07)
Gait Velocity (m/min)	66.57 (5.28)	63.70 (6.48)	67.42 (5.52)	65.38 (5.34)
Cadence (steps/min)	101.55 (4.95)	102.75 (4.52)	101.22 (4.27)	102.39 (6.03)
Stance(% stance)	0.61 (0.03) *	0.65 (0.02) **	0.63 (0.02)	0.64 (0.02)
Stance Time (s)	0.73 (0.04)	0.76 (0.04)	0.75 (0.03)	0.75 (0.034)

4.3 Gait Curves

Gait curves for each of the five subjects wearing both the NPO and SAFE foot were graphed. Gait curves were created for five different measures: Forces, Moments, Powers, Angles, and GRF. Each measure was separated by leg (residual, sound) and by axis (Posterior-Anterior, Medio-Lateral, Distal-Proximal. All graphs are presented in Appendix G.

Graphs of the five significantly curve parameters (see 4.4 Curve Parameters) that differ between prosthetic feet for the residual limb are presented below (see figures 4.1 – 4.10). There were no gait pattern differences on the sound limbs between either prosthetic foot. Graphs for the sound limb gait curves can be seen in Appendix G.

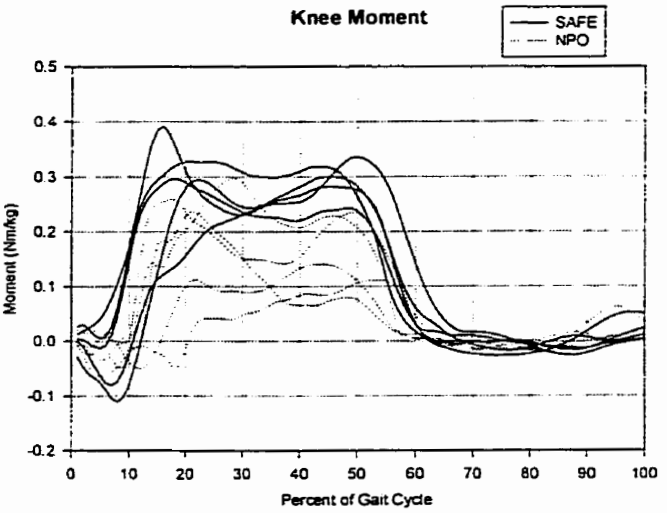


Figure 4.1: Average Fixed Body Moment at the Knee Joint - Residual Limb
Five subjects, average for five trials per subject, by prosthetic, Posterior-Anterior

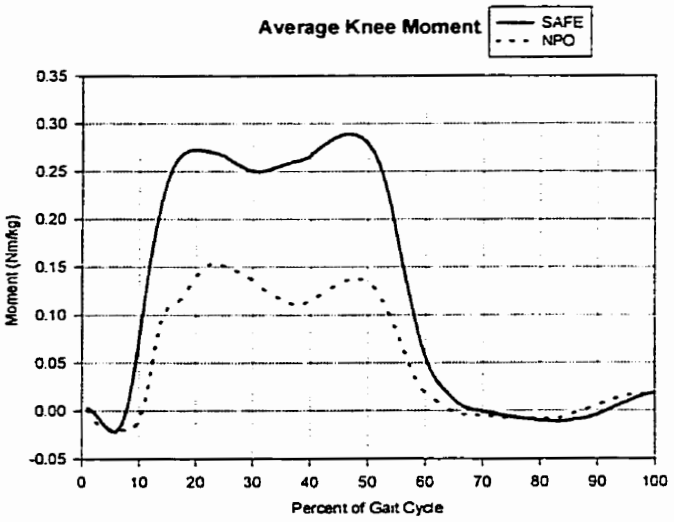


Figure 4.2: Average Fixed Body Moment at the Knee Joint - Residual Limb
Five subjects, averaged by prosthetic, Posterior-Anterior

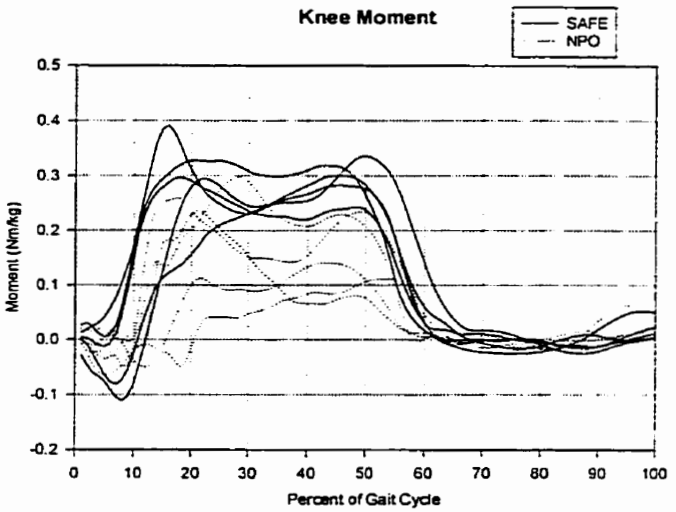


Figure 4.3: Average Fixed Body Moment at the Knee Joint - Residual Limb
Five subjects, average for five trials per subject, by prosthetic, Medio-Lateral

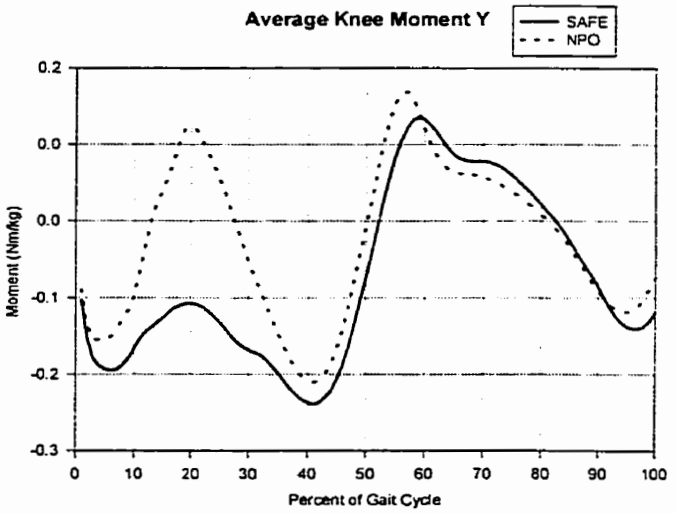


Figure 4.4: Average Fixed Body Moment at the Knee Joint - Residual Limb
Five subjects, averaged by prosthetic, Medio-Lateral

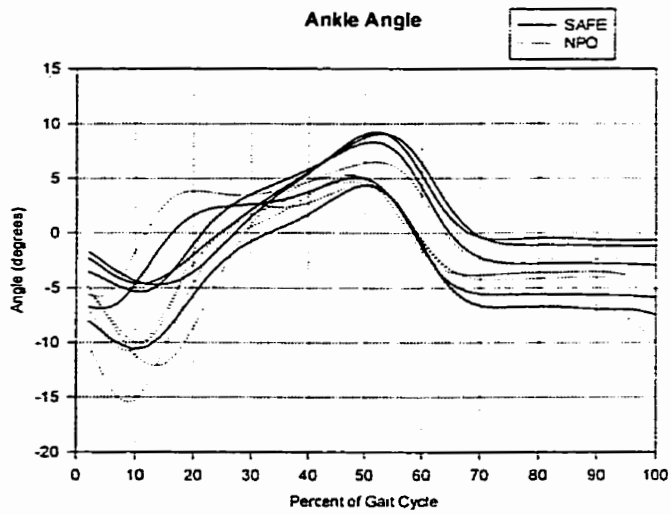


Figure 4.5: Average angle at the ankle joint - Residual Limb
Five subjects, average for five trials per subject, by prosthetic, Medio-Lateral

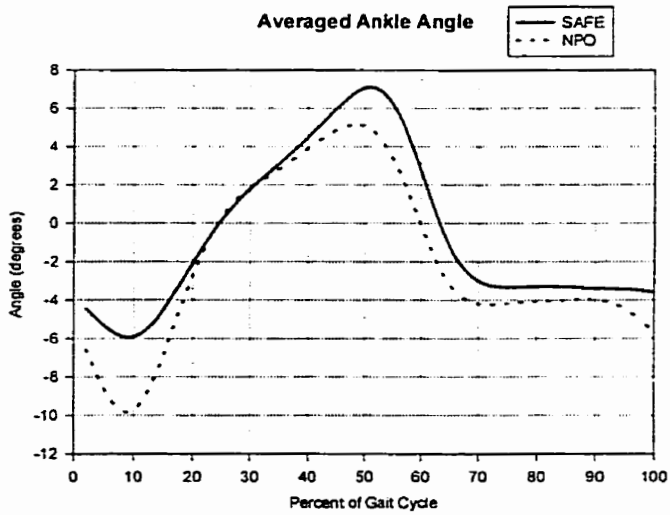


Figure 4.6: Average Angle at the Ankle Joint - Residual Limb
Five subjects, averaged by prosthetic, Medio-Lateral

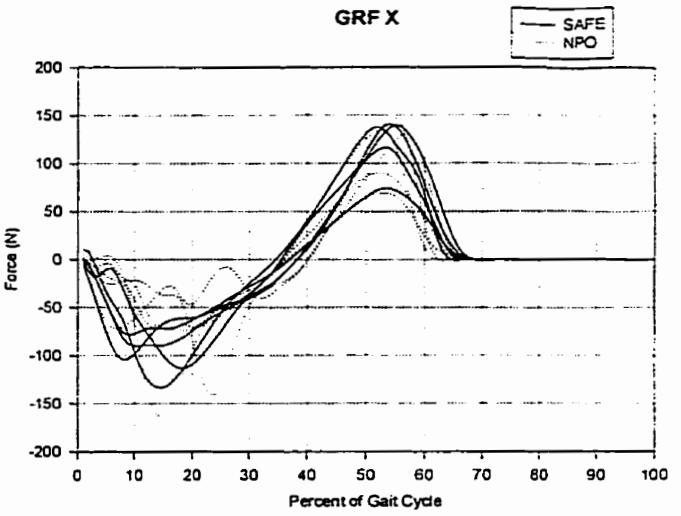


Figure 4.7: Average ground reaction force - Residual Limb
Five subjects, averaged for five trials per subject by prosthetic, Posterior-Anterior

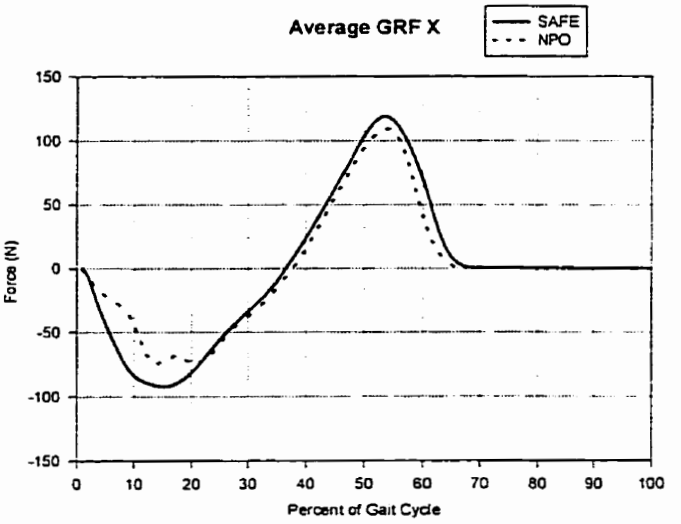


Figure 4.8: Average ground reaction force - Residual Limb
Five subjects, averaged by prosthetic, Posterior-Anterior

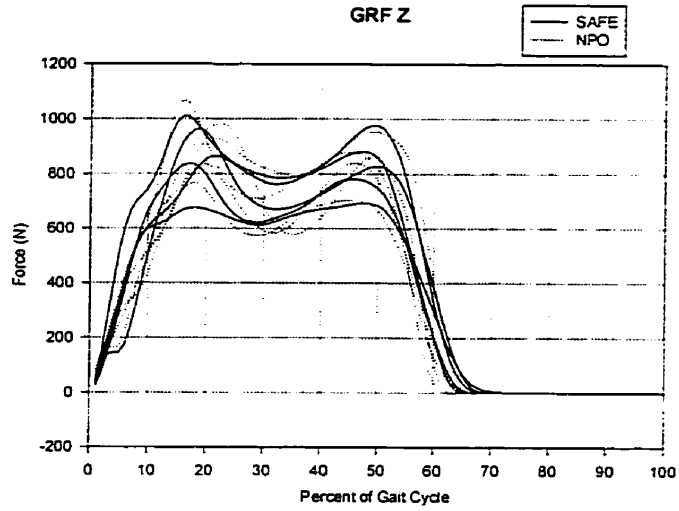


Figure 4.9: Average ground reaction force - Residual Limb
Five subjects, averaged for five trials per subject by prosthetic, Distal-Proximal

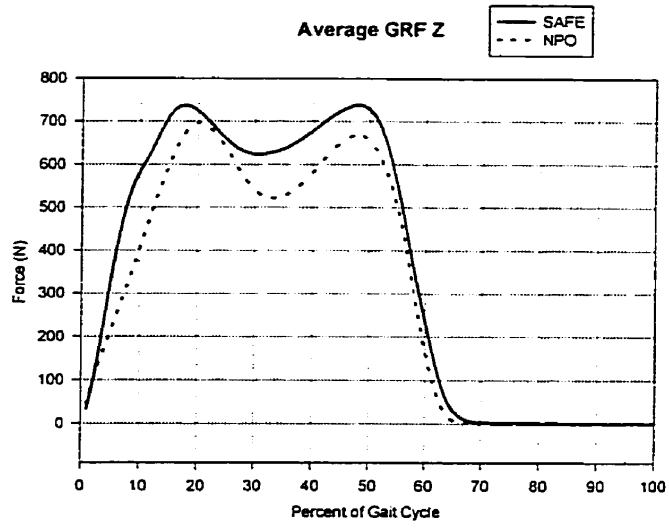


Figure 4.10: Average ground reaction force - Residual Limb
Five subjects, averaged by prosthetic, Distal-Proximal

4.4 Curve Parameters

Thirty four events from the gait curves (forces, moments, powers, angles, and GRF) were chosen as variables to compare the gait profiles produced by the two prosthetic feet. All gait curve graphs are presented in Appendix G.

All five trials for each subject were graphed allowing a visual observation of the data to determine the existence of outliers. Curve parameter values for all subjects in both the NPO and SAFE prosthetic feet are presented in Appendix H.

Variable names for the curve parameter were constructed in four sections to indicate **Measure, Joint, Gait Curve, and Plane**. The following convention was use to create the curve parameter variable names:

Measure:

Max – Maximum

Min – Minimum

Rng – Range

Lmax – Location of Maximum

Lmin – Location of Minimum

Slp – Slope

+Area – Positive Area under the curve (impulse)

-Area – Negative Area under the curve (impulse)

ZX – Location of zero crossing

Joint (if not is listed, Knee joint is default)

K – Knee joint

A – Ankle Joint

Gait Curve

FF – Fixed Force

FM – Fixed Moment

A – Angle

P – Power

FP – Force Plate (GRF)

Axis

X – Posterior-Anterior

Y – Medio-Lateral

Z – Distal-Proximal

Example: LmaxKFMX – Location (L) (as a % of gait cycle) of the maximum (max) knee (K) fixed body (F) moment (M) about the Posterior-Anterior axis (x) axis.

Curve parameter data were averaged across five trials to obtain subject average values. These values averaged to provide an average for each prosthesis (NPO and SAFE). The means and standard deviations for all 34 curve parameters for the residual limb are presented in Table 4.3 through 4.7. Significant differences were found for six of the 34 variables. Differences were not detected in the remaining 28 variables.

Two differences occurred in curve parameters from the knee moment curves. The maximum moment at the knee joint about the LM axis between 40 and 100 percent of the gait cycle was found to be smaller for the NPO foot. The average peak moments for the NPO and SAFE feet were found to be 0.17 Nm/kg (± 0.07), and 0.30 Nm/kg (± 0.05) respectively. The location of the maximum moment about the PA during the gait cycle was also found to occur earlier in the NPO foot compared to the SAFE prosthetic. The average location of maximum moment was 38.18 % of gait cycle (±19.29) for the NPO prosthetic and 57.54 % of gait cycle (±9.95) for the SAFE (see table 4.3.)

One different curve parameter was found in the joint angle data. The minimum ankle angle (plantarflexion angle) of the NPO foot was larger than that of the SAFE foot. The values of for the NPO and SAFE foot were –10.98 degrees (±3.28) and –6.47 degrees (±2.27) respectively (see table 4.6)

Four different curve parameters were found in the ground reaction force data. The NPO prosthesis had a smaller impulse between heel strike and foot flat than did the SAFE foot with values of 19.02 Ns (±3.77) and 22.40 Ns (±4.23) respectively. A similar relationship was found in the impulse created between foot flat and toe off. The means for the NPO foot and SAFE feet were -19.39 Ns (±3.87) and –23.14 Ns (±2.73) respectively. Maximum vertical ground reaction force between 0 and 30 percent of the gait cycle was the third different GRF curve parameter. The averaged value for the NPO foot was 920.55 (±111.08) N which was larger than the 873.61 (±121.49) N for the SAFE foot (see Table 4.7). The slope of the vertical GRF after hell strike was smaller compared to the SAFE foot, with values of 52.18 (±9.84), and 65.30 (±9.96) respectively.

Table 4.3: Force curve parameters averaged for five subjects by prosthetic

Variable Name	Location in Gait Cycle (%)	NPO (N/kg)	SAFE (N/kg)
MinKFFX	0-20	-1.32 (0.24)	-0.87 (0.24)
MaxKFFX	0-100	2.62 (0.78)	2.31 (0.32)
MinKFFY	0-100	-0.80 (0.24)	-0.94 (0.22)
MinKFFZ	0-100	-10.22 (0.54)	-9.84 (0.54)

* - Significant difference (alpha = 0.05)

Table 4.4: Moment curve parameters averaged for five subjects by prosthetic

Variable Name	Location in Gait Cycle (%)	NPO (Nm/kg)	SAFE (Nm/kg)
MaxKFMX	0-40	0.22 (0.09)	0.32 (0.06)
MaxKFMX	40-100	0.17 (0.07) *	0.30 (0.05) *
LMaxKFMX	0-40	23.56 (10.20)	24.28 (8.79)
LMaxKFMX	40-100	52.51 (2.52)	53.15 (1.87)
MaxKFMY	0-100	0.30 (0.16)	0.18 (0.08)
LMaxKFMY	0-100	38.18 (19.29) *	57.54 (9.96) *
RngKFMY	0-100	0.72 (0.17)	0.53 (0.12)
MaxKFMZ	0-100	0.06 (0.03)	0.08 (0.02)

* - Significant difference (alpha = 0.05)

Table 4.5: Power curve parameters averaged for five subjects by prosthetic

Variable Name	Location in Gait Cycle (%)	NPO (Nm/s/kg)	SAFE (Nm/s/kg)
MaxKPX	0-20	0.10 (0.09)	0.12 (0.08)
MaxKPX	20-100	0.07 (0.09)	0.10 (0.10)
MaxKPY	0-100	1.14 (0.52)	0.97 (0.20)
MinKPY	0-100	-0.48 (0.28)	-0.32 (0.13)
MaxKPZ	0-100	0.03 (0.01)	0.05 (0.02)
MinKPZ	0-100	-0.03 (0.02)	-0.03 (0.02)

* - Significant difference (alpha = 0.05)

Table 4.6: Angle curve parameters averaged for five subjects by prosthetic

Variable Name	Location in Gait Cycle (%)	NPO (degs)	SAFE (degs)
RngKAX	0-100	11.16 (6.60)	15.05 (6.98)
MaxKAY	0-50	16.60 (3.65)	10.10 (6.21)
MaxKAY	50-100	65.05 (3.01)	68.41 (3.99)
RngKAZ	0-100	13.85 (3.87)	15.01 (4.17)
MinAAX	0-20	-10.90 (3.28) *	-6.47 (2.27) *
MaxAAX	10-60	5.32 (0.74)	7.27 (2.18)
RngAAX	0-100	16.34 (3.58)	13.73 (1.09)

* - Significant difference (alpha = 0.05)

Table 4.7: Ground reaction force curve parameters for five subjects averaged by prosthetic

Variable Name	Location in Gait Cycle (%)	NPO (N)	SAFE (N)
+AreaFPX	0-100	19.02 (3.77) *	22.40 (4.23) *
-AreaFPX	0-100	-19.39 (3.87) *	-23.16 (2.73) *
ZXFPX	20-50	36.40 (1.98)	37.20 (2.60)
MaxFPY	0-30	62.39 (14.66)	54.26 (21.77)
MaxFPY	30-60	52.62 (16.09)	46.37 (20.28)
MaxFPZ	0-30	920.55 (111.08) *	873.61 (121.49) *
MaxFPZ	30-60	860.26 (77.48)	835.20 (98.25)
MinFPZ	20-50	638.69 (86.52)	687.09 (72.59)
SlpFPZ	0-11	52.18 (9.84) *	65.30 (9.96) *

* Significant difference (alpha = 0.05)

4.5 Subjective Feedback

Subjective feedback from the subjects in the subjective questionnaire was summed. The questionnaire form can be seen in Appendix D. The questions asked were as follows:

1. Comfort of the prosthetic
2. Ease of use
3. Ease of adaptation
4. Stability when standing
5. Stability when walking

6. Minimizes muscular effort
7. Heel strike feels good
8. Toe off feels good
9. Opposite leg feels good
10. Limb/Socket contact feels good

Results from the subjective feedback questionnaire can be seen in table 4.8.

Table 4.8: Questionnaire results from five subjects

	CL		DK		MA		MM		SG	
Question	NPO	SAFE	NPO	SAFE	NPO	SAFE	NPO	SAFE	NPO	SAFE
1	6	7	7	10	9	8	10	10	9	8
2	6	8	8	10	9	8	7	10	8	7
3	7	9	8	10	9	8	9	10	8	5
4	9	9	9	10	9	10	10	9	8	6
5	9	8	9	9	9	9	8	10	9	6
6	8	7	8	9	9	7	9	10	8	6
7	6	6	7	9	10	7	6	10	6	3
8	4	6	7	10	9	8	6	10	8	4
9	6	8	8	9	10	7	10	10	9	9
10	8	9	7	9	5	8	10	10	9	8
	NPO	SAFE	NPO	SAFE	NPO	SAFE	NPO	SAFE	NPO	SAFE
Total Score	69	77	78	95	88	80	85	99	82	62

When the scores for each of the ten questions were compared there was no agreement across all five subjects as to which foot was superior. Based on these total scores two subjects (MA and SG) perceived the NPO foot to have overall preferred performance, while the remaining three perceived the SAFE's performance to be superior.

Subjects were asked to comment on their overall perceptions of the NPO foot's performance. All five subjects indicated that they were pleased with NPO's performance. Common feedback included an indication that the NPO foot produced a somewhat hard heel strike and a mediolateral "kick" or "whip" (the prosthesis would turn quickly, twisting the foot inward or outward upon heel strike). Subjects MM, DK, and SG all reported playing sports while wearing both feet. These subjects reported that the NPO prosthesis was preferable for these activities.

5.0 Discussion

5.1 General

Before a new prosthetic foot can be introduced to the general public, some testing must be performed to assure that the gait patterns are not so abnormal as to cause pain or injury. Quite often the testing of new prosthetic foot involves only subjective evaluation from a small to medium size sample of BK amputees. Certain kinetic and kinematic properties of gait patterns are not evident from visual inspection or subject perceptions. At present no set standards are available to determine what levels of forces, moments, and powers at the joints are acceptable. However, it is felt that large deviations in forces or moments can affect the joints adversely (Perry 1975)(Brouwer, Allard, et al. 1989)(Lewallen, Dyck, et al. 1986)(Marks, Palmer, et al. 1978). This is what makes comparison studies with the SACH foot so valuable, as it provides a "normal" or "baseline" set of gait patterns. The SACH foot is the industry standard and both the scientific and clinical communities have deemed its well studied gait patterns acceptable.

The primary goal of a comparison based prosthetic gait study is to determine differences in gait patterns between a new prosthetic foot and a comparison foot. This is essential in determining the suitability of a new prosthetic foot (Seliktar & Mizrahi 1986). If the gait patterns of the two feet are indistinguishable then the performance of the new prosthetic foot can be assumed to be as good as that of the comparison foot. If differences were found the goal would then be to establish whether the differences are problematic (Seliktar & Mizrahi 1986). For example, problems such as the production of excessively large knee moments may harm the patient, rending the component unusable. Viewing overall gait pattern changes between the feet identifies areas of difference. Measuring events in those patterns enables the researcher to quantify the differences that can lead to problems.

A secondary goal is to identify properties of the prosthetic foot that affect efficiency of gait. For example, poor braking or power generation while walking could reduce the performance or efficiency of gait. When kinetic property differences are found kinematic data (e.g. joint angles) can shed light on foot mechanics that may cause gait deficiencies.

These results are valuable in the development and redesign of a new prosthetic foot (Lehmann, Price, et al. 1993).

The effect a prosthetic foot has on the contralateral limb must also be considered. Deficiencies in the performance of the residual limb may result in the subject adopting compensatory movement patterns in the sound limb. For example, the inability to properly clear the toe in the swing phase on the residual limb may be compensated for by circumduction of the contralateral hip. Increased loading or reliance on the joints of the sound limb is an indication of a poor prosthetic foot (Eberhart 1968) and increased loading could lead to accelerated joint degeneration (Marks, Palmer, et al. 1978)(Seliktar & Mizrahi 1986).

When no difference is found then the performance of the two feet is assumed to be comparable. Detection of differences will indicate performance differences in the feet. Quantification of these differences and their possible consequences for the amputee may be used to draw a conclusion as to whether the new prosthetic foot is suitable for general use.

5.2 Mechanical Operation of Prosthetic Feet

The two prosthetic feet that were tested used different mechanical mechanisms to replace gait function lost by amputation. Below is a general description of the mechanical operation the feet.

5.2.1 SAFE Foot Mechanics

The SAFE prosthetic foot like other SACH type feet has a soft, sponge-like material in the heel. During heel strike, when the individual begins weight bearing on the residual limb, the cushioned heel of the prosthetic foot compresses. This serves two main functions. First, the cushioned heel alters both the vertical and horizontal impulses produced during forward slowing of the body's center of mass at heel strike. The result is vertical shock absorption and a reduction in the peak force transfer to the stump and

joints above the stump. Absorption of the forward (horizontal) force creates a braking impulse (an impulse in opposition to the direction of progression).

Secondly, the solid ankle of the SAFE foot does not allow the foot to plantar or dorsiflex as would a normal articulated ankle joint. The compressible heel of the SAFE foot simulates the normal plantarflexion function created by the articulating ankle joint. As the heel is compressed the foot sinks toward the ground allowing the individual to achieve a foot flat stance.

The solid ankle of the foot also prevents true dorsiflexion from occurring. Unlike its predecessor (the SACH) the SAFE foot does not possess a completely rigid keel. The foot has a flexion point corresponding to the metatarsal-phalangeal joint of the normal foot. This aids in the transfer from foot flat to toe off by simulating dorsiflexion.

5.2.2 NPO Foot Mechanics

The NPO prosthetic foot can best be categorized as a rolling joint foot. The "S" shape of the foot-ankle complex creates a joint with an axis of rotation that changes location during the stance phase as the body's center of mass proceeds forward over the foot. The "S" shaped ankle joint can be described as two "C" curves, one on top of the other, facing opposite directions. At heel strike the top C curve opens and the lower curve compresses allowing plantarflexion to occur. This action alters the vertical and horizontal impulse allowing weight acceptance. The C curves also let the base of the foot rotate towards the floor (plantarflex). As the subject achieves foot flat and begins to move over the foot the top C curve begins to close during weight acceptance. At the end of mid stance the individual's center of mass begins to move toward the front of the prosthetic foot. At this point the top C curve closes completely. As the center of mass continues forward the bottom C opens to created dorsiflexion allowing the shank to rotate forward with respect to the foot.

The NPO foot is made of a flexible material that is thinner in the forefoot than in the hindfoot. This allows the forefoot of the prosthetic foot to bend. During mid stance the top C curve closes completely and touches the top of the forefoot section. The vertical

forces are then transferred to forefoot section which bends at a location corresponding to the metatarsal-phalangeal joint.

The material chosen for the NPO foot was a polyethylene compound. The properties of this material allow the foot to be strong and durable, yet flexible. The strength of the material prevents the foot from breaking while altering the vertical and horizontal impulses during weight bearing. Its high durability prevents breakdown due to repeated deformations of the material during gait. Although the material is strong, and durable it is supple enough to allow the "S" ankle joint to flex properly. The combination of these attributes makes the polyethylene material an important part of the foot design.

5.3 Time Distance Parameters

Time distance parameters are the most commonly reported quantitative measures reported in amputee gait studies. Although the reasons for a performance deficiency may not be evident from the observation of time distance parameters, these stride characteristics can indicate overall performance differences between prosthetic feet.

No differences were detected for the time distance parameters on the sound limb. However, data from the residual limb showed that the stance ratio was significantly smaller for the NPO foot at 0.61 compared to 0.65 ($p= 0.009$) for the SAFE foot (see table 4.2). The value of 0.65 for the SAFE prosthetic foot is consistent with other studies that tested the stance ratio of this foot. Previous studies have reported stance ratios from 0.62 – 0.66 (Doane & Holt 1983)(Culham, Peat, et al. 1986). This suggests that subjects spent less time in stance phase while wearing the NPO foot. The reduced time spent in stance phase for the NPO foot could be due to an unconscious effort by the amputee to shift weight more quickly from the residual limb to the sound limb. This quick shift could indicate a lack of confidence in the stability of NPO foot or reduced comfort during weight bearing (Bateni 1996)(Culham, Peat, et al. 1986) (Suzuki 1972). Questions 1,4,5, and 10 from the post trial questionnaire evaluated the subjects' perceptions of the stability and comfort. Feedback did not consistently support either of these hypothesizes.

It has been shown that amputees have different stance ratios than able-bodied individuals (Robinson, Smidt, et al. 1977). Able-bodied individuals normally exhibit stance ratios of 0.6 – 0.62, spending 60 - 62% of total gait time in stance phase and 38 - 40% in swing phase and this ratio is constant for both limbs (Rose & Gamble 1996). Below knee amputees exhibit different stance ratios for their residual and sound limbs (Culham, Peat, et al. 1986)(Prince, Allard, et al. 1992). Amputees often spend a higher than normal percentage of total gait time in stance phase on their sound limb and a lower than normal percentage of total gait time in stance phase on the residual side (Torburn, Perry, et al. 1990). This is referred to as an asymmetric gait. The larger the difference between the two limbs the more asymmetrical the gait pattern becomes and the less efficient (Hurley, McKenney, et al. 1990).

A greater amount of symmetry than expected was found between the residual and sound limbs for both prosthetic feet. Stance percentages for the sound limbs have been found to be as large as 0.65 – 0.71 in other studies (Culham, Peat, et al. 1986)(Hurley, McKenney, et al. 1990)(Lewallen, Dyck, et al. 1986). Results from this study showed that sound limb stance percentages were much closer to their associated residual limb levels for both feet, suggesting more symmetry than previously reported.

Time distance parameters correlate highly with each other (Robinson, Smidt, et al. 1977) and differences in stance ratios are often accompanied by different stance times. Stance time while wearing the NPO prosthetic foot (0.725seconds) appeared to be slightly shorter than that of the SAFE prosthetic at (0.757 seconds) which would be expected given their associated stance ratios. However data analysis revealed no statistical difference between the prosthetics ($p = 0.088$).

Gait velocity and step length are two time distance variables known to correlate very highly. Values for both variables have been shown to be lower in amputees compared to able-bodied individuals and are excellent indicators gait dysfunction (Robinson, Smidt, et al. 1977). Significant reduction in these variables indicates a less effective gait pattern and may be a strategy used by amputees to reduce loading on uncomfortable or unstable prosthetics (Bateni 1996)(Culham, Peat, et al. 1986). Changes in velocity and step length can cause alterations in the stance ratio. The gait velocity and stride length of both the NPO and Safe foot were less than values associated with able-bodied gait. However,

values for both feet did fall within the range considered normal for BK amputees' (see table 5.1) (Barth, Schummacher, et al. 1992)(Culham, Peat, et al. 1986) (Doane & Holt 1983)(Lemaire, Fisher, et al. 1993)(Robinson, Smidt, et al. 1977)(Torburn, Perry, et al. 1990).

Table 5.1 Comparison of average gait velocity and stride length from seven amputee gait studies

Study	Gait Velocity (m/min)	Stride Length (m)
Barth et al. (1992)	45	1.1
Culham et al. (1984)	34.37	0.941
Doane and Holt (1974)	73.2	N/A
Lemaire et al. (1993)	72	1.41
Robinson et al. (1977)	64.2	1.32
Synder et al. (1995)	63.3	1.25
Torburn et al. (1990)	70.2	1.4
This study (NPO) (SAFE)	(66.57) (63.70)	(1.31) (1.24)

No difference could be detected between the gait velocity ($p = 0.172$) or step length ($p = 0.057$) variables for the feet. There was also no difference detected for the stride time ($p = 0.37$) or cadence ($p = 0.40$) between feet, suggesting that the overall performance of the feet with regard to these variables was comparable.

The cumulative effect of both advantageous and disadvantageous kinematic and kinetic properties are reflected in time distance parameters (Collins & Whittle 1989). That is why they are an excellent indicator of overall performance differences. The inability to detect difference in many of these variables for both residual and sound limbs suggests that the overall performance of the NPO foot is comparable to the SAFE foot. Decreases in the stance ratio may indicate a deficiency in performance cause by an individual's perception of less weight bearing stability or a general lack of confidence in the NPO foot. The small sample size ($N = 5$) and the lower observed power from several tests should be considered when drawing conclusion based on these variables.

5.4 Gait Curves

Although empirical means of assessing the performance of prosthetic feet is the focus of this study, the contribution from clinical observations should not be discounted as they provide important information that may be masked by viewing only the selected events within the gait curves. Visual inspection of the gait curves can be used to observe major pattern deviations or major differences between patterns produced by different feet (Menard, McBride, et al. 1992) (Seliktar & Mizrahi 1986). Although the differences in patterns pointed out in the following discussion are not quantified, they do suggest possible important functional differences between the feet.

When the gait curves from the sound limb were compared no major differences were found between the patterns created when wearing the NPO prosthetic foot versus the Safe prosthetic foot. Gait curve patterns for forces, moments, angles, powers, and GRF were all similar to each other and the NPO foot exhibited no major deviations from SAFE foot gait patterns (see Appendix G).

Inspection of the gait curves produced by the NPO and SAFE foot for the residual limb reveled several notable differences. The gait patterns produced by both prosthetic feet appeared to exhibit more variability in the gait patterns of able-bodied individuals which is consistent with previous research (Lemaire, Fisher, et al. 1993)(Seliktar & Mizrahi 1986) (Smidt 1990). Gait curves from this study appeared more variable both within trials and between subjects than would be expected from able-bodied individuals. This increased variability both between trials and between subjects can be attributed to variations in the residual limbs. Both the length of the stump, and the stability of its volume can add to the variability seen between individuals (Robinson, Smidt, et al. 1977).

Pattern differences in GRF data from the Posterior-Anterior plane (fore aft), residual limb suggests that the NPO does not produce a normal braking force (figure 4.3, figure G9). The gait pattern from SAFE foot appears to be more consistent between trials and between subjects than that of the NPO. The SAFE braking impulse pattern also more closely resembles that of able-bodied individuals. The less consistent pattern created by the NPO foot is likely due to its square heel and the lack of cushioning. The rounded, cushioned heel of the SAFE prosthetic foot provides a more even stable braking force at heel strike. Its rounder configuration is more forgiving of slight angular or rotational deviation of the foot at heel strike.

The square heel and lack of cushioning in the heel of the NPO foot likely causes the foot to twist or rotate when the foot is not perfectly straight at heel strike. When heel contact is made, force is transferred through the heel to the ground, creating a force opposite in direction to the direction of walking. If the foot is twisting or rotating at initial contact with the ground, some force will be lost to this rotation, and lessen the force used to create the braking impulse. This would account for its abnormal braking impulse reflected in the GRF data. This is supported by the subjects who commented that the NPO foot 1) seemed harder and less comfortable at heel strike and 2) had a tendency to "kick" or twist medially at heel strike. Prior to data processing which included data filtering, more high frequency noise was noted in the gait trials from the NPO foot at the knee and hip joints. This can also be attributed to the hard surface of the NPO foot that appears to have a less ability to act as a shock absorber.

A small difference in residual limb ankle angle pattern was noted (see figure 4.3, G15). This difference is not unexpected given the different mechanisms by which the two feet simulate the functions of the articulated ankle joint. The SAFE foot demonstrates a slightly smoother single peak pattern. The NPO foot appears to have a two peak pattern. This is most likely due to the transfer of the center of rotation from the top to the bottom curve of the NPO's "S" shaped ankle (see section 5.2.1, 5.2.2).

Differences were also noted in the forefoot angles of the residual limb (see figure G17). The break in the keel of the SAFE foot appears to allow a slightly higher degree of forefoot flexion prior to toe off. This pattern appears to be considerably more variable than the consistent forefoot action of the NPO prosthetic foot. The action of the NPO foot is dependent on the flexibility of the material and it thickness. A decreased forefoot angle prior to toe off suggests the need for more flexible material or a design change to thin the material at the forefoot allowing increased flexion.

5.5 Curve Parameters

A comparison of all curve parameters derived from the gait curve of the sound limb revealed that no significant differences could be detected in any of the 34 variables. This suggests that gait produced by NPO prosthetic foot does not adversely affect the movement patterns of the sound limb and no major compensatory changes from SAFE foot gait patterns are created.

Six curve parameters for the residual limb were found to be different between feet. Two parameters from the joint moment curves and four parameters from the ground reaction force curves. No differences could be detected between the remaining twenty eight parameters. The performance of the feet based on those variables is assumed to comparable.

5.5.1 Knee moments

Ground reaction forces produced during gait are transferred through prosthetic foot to the intact knee joint of the amputee. Below knee amputees no longer possess an ankle joint or the muscles of the shank that allow able-bodied individuals to more readily control ground reaction impulse. The lack of shock absorption increases the deviations of prosthetic gait from able-bodied gait and also decreases the amputee's ability to accommodate to gait perturbances. The amputee must therefore rely on the prosthetic foot to minimize the effects of the force transferd from the ground to the knee joint. It is believed that any alteration of the force and moment patterns at the knee joint greatly increases the possibility of osteoarthritis in the knee joints, particular in the residual limb. For this reason the magnitude and patterns of the forces and moments experienced at the knee joint must be carefully examined before introducing a new prosthetic foot.

The magnitude of the maximum moment experienced at the knee (flexion / extension) was smaller while wearing the NPO foot compared to the SAFE foot (figure 4.1,4.2). The average maximum knee flexion moments for the NPO and SAFE prosthetic feet were 0.17 Nm/kg and 0.30 Nm/kg ($p = 0.004$) respectively. These magnitudes are within the range found by other studies. Moments for the subjects were lower than values for able-bodied gait regardless of which prosthetic foot was worn, supporting the findings of the earlier research (Smidt 1990). The values attained for the SAFE prosthetic foot are consistent with moments reported in previous studies using that prosthetic foot (Mueller, Minor, et al. 1995)(Czerniecki, Gitter, et al. 1991). The difference in the magnitude of peak knee moments (PA axis) can be explained by examining the mechanics of the two feet. The solid ankle of the SAFE foot does not allow the shank to rotate with respect to the foot. The break in the keel of the SAFE prosthetic foot does allow some minimal flexion of the forefoot (simulating the metatarsal-phalangeal break) just prior to toe off. During the time after full compression of the cushioned heel, but prior to the toe off, when there is a shift from plantarflexion to dorsiflexion, the solid ankle joint of the SAFE foot is inflexible. The forward progression of the center of mass over the now fixed joint of the SAFE has been shown to create a large internal dorsiflexion moment at the ankle

that is transferred through the shank and affects peak moments at the knee joint (Winter & Sienko 1988). In contrast, the rolling joint of the NPO foot flexes throughout the entire stance phase. The two curve "S" design allows easy progression from plantarflexion to dorsiflexion. This ability to flex throughout the stance phase more closely resembles the motion of an articulated joint. By allowing rotation of the shank throughout the stance phase the NPO produces a smaller dorsiflexion moment at the ankle resulting in a lower peak knee moment.

The timing of the peak knee moment about the Medio-Lateral axis differed between the two feet, occurring at 38.2% of gait cycle for the NPO foot, and 57.5% of gait cycle for the SAFE foot ($p = 0.014$) (figure 4.2, 4.3). This suggests that the peak knee moment for the NPO foot occurs when the line of gravity is being transferred from the heel to a position in line with the shank. In contrast, the peak knee moment of the SAFE prosthetic foot occurs immediately prior to toe off. This difference is likely due to differences in the shape of the bottom of the prosthetic feet. The bottom of the SAFE prosthetic foot is contoured to simulate the arches of a real foot and allows easier medio-lateral rolling of the foot during gait. In contrast, the NPO prosthetic foot is flat and uniform across the bottom. This makes rolling of the foot medio-laterally difficult and causing higher abduction-adduction moments at the knee following heel strike.

5.5.2 Ankle Angle

The NPO foot had a larger degree of plantarflexion near heel strike exhibiting approximately four degrees more plantarflexion than the SAFE foot. The plantarflexion produced by the NPO prosthetic foot is also a more "true" plantarflexion. Plantarflexion produced by the SAFE prosthetic foot is simulated plantarflexion because the keel of the foot does not rotate towards the ground. With its solid ankle the position of the shank in reference to the prosthetic foot is constant. To allow an individual to achieve a foot flat position the heel of the SAFE compresses when the vertical force of weight bearing is applied to it. This allows the foot to "sink" towards the ground bringing the individual to foot flat. The sponge material in the heel of the SAFE prosthetic foot does not return the

vertical force following heel strike to help accelerate the body upward. The disadvantage of simulating plantarflexion in this way is that the residual limb is displaced vertically downward the distance that the heel compresses. With no significant energy return from the heel (vertically), additional energy is required from the sound limb to raise the residual limb back up. Prosthetists have referred to this phenomenon as the SACH "hole". In contrast, when plantarflexion is created with the NPO foot the vertical position of the residual limb is unchanged. This advantage eliminates the SACH "hole" and may minimize metabolic costs. The NPO foot and residual limb also rotate in a pattern that more closely mimics the natural gait produced by an articulating ankle.

5.5.3 Ground Reaction Forces

Parameters derived from GRF data are a valuable means of assessing overall gait performance by providing information on the magnitude and direction of the forces acting on the body during gait. Prior to heel strike, the body's center of mass is accelerating forward and moving outside the base of support. At heel strike force produced by the muscles of the lower limbs is applied to the ground through the foot to create an impulse in the direction opposite to movement to the center of mass (Winter 1991). This impulse causes a deceleration the body's center of mass in the fore-aft direction and is a "braking" impulse. Prior to toe off, force created by leg muscle contraction applied through the foot to the ground creates an impulse in the direction of forward progression. This impulse is referred to as "propulsive" impulse and accelerates the body forward (Winter 1991). Observing the magnitude of these impulses allows the determination of how effectively a prosthetic foot transfers force produced by the musculature of the residual limb to the ground to create forward momentum in gait. Decreases in either the braking or propulsion impulse indicate a less efficient transfer of force and therefore a less efficient gait (Seliktar & Mizrahi 1986).

The braking impulse created by the NPO prosthetic foot was found to be smaller than that produced by the SAFE prosthetic foot ($p = 0.022$) (see table 4.7). The average impulse of the NPO and SAFE feet was –19.39 N/s and –23.16 N/s respectively. The NPO foot was also found to create a smaller propulsive impulse than the SAFE prosthetic

foot ($p = 0.010$) (see table 4.7). The average impulse of the NPO and SAFE feet were 19.01 N/s and 22.34 N/s respectively. The SAFE produce a larger propulsive impulse than the NPO foot. This increased impulse creates forward momentum to drive the body forward.

The smoother, more even braking gait pattern of the SAFE foot (see 5.3) appears to provide stable, balanced heel contact. This reflects a more stable prosthetic that is less likely to caused perturbation in the gait patterns of the lower limb joints (Seliktar & Mizrahi 1986).

A significantly larger maximum peak in vertical GRF was found for the gait produced in the NPO prosthetic foot. The mean maximum peak vertical ground reaction force for the NPO and SAFE feet were found to be 920.55 N and 873.61 N ($p = 0.044$). The NPO foot did allow a higher degree of dorsiflexion at heel strike yet produced a higher vertical GRF peak (at gait cycle 0-11%). This increase is most likely due to the NPO foot's lack of cushioning. The sponge-like heel of the SAFE prosthetic foot act as a shock absorber during heel strike. With no cushioning at the heel, or spongy cosmetic coating, the hard plastic surface of the NPO prosthetic foot is responsible for creating an increased vertical GRF peak during heel strike.

Several gait characteristics can lead to increases in the vertical GRF. These include increases in gait velocity, stride length, decreases in knee flexion angles at heel strike, and decreased muscular activity in the muscles of the lower limb (Collins & Whittle 1989) (Lewallen, Dyck, et al. 1986) (Sanderson & Martin 1996). No detectable differences were found between the feet for gait velocity, stride length, or knee angles (muscle activity was not monitored in this study). We are therefore confident in concluding that the differences in GRF are due solely to the feet.

The NPO prosthetic foot produced a smaller vertical GRF slope than the SAFE prosthetic foot. This slope is calculated as the change in force after heel strike compared to the change in time over the first 12% of the gait cycle. Visual inspection of the vertical GRF gait curves for the residual limb (figure G9) revealed that the maximum peak for NPO foot occurs latter in the gait cycle than the maximum peak for the SAFE foot. Although the NPO foot produces a larger vertical GRF its design allows it to modify the impulse over a longer time period, thus reducing the slope. The rolling "S"

joint of the NPO prosthetic foot provides a longer period over which the vertical GRF is changed. This is most likely due to its increased ability to produce a higher degree of the ankle plantar flexion. A second possibility is provided by Seliktar et. al (1986) which suggests that moderation of the slope may be evidence of caution by the amputee and represents a less stable foot ankle complex.

The slope of the vertical GRF during heel strike (0-11% gait cycle) provides information on the rate at which the total vertical GRF is absorbed by the residual limb. The SAFE foot's smaller maximum peak forces absorb more quickly (higher slope) may cause more problems to the knee joint than the NPO's larger force absorbed at more gradual rate. One advantage that a faster deceleration may provide is a reduced time to foot flat.

5.6 Summary

In summary, significant differences were detected for the following eight variables:

1) Stance ratio
2) Maximum knee moment about the Lateral-Medial axis (40-100% of GC)
3) Location of the max knee moment Posterior-Anterior axis (0-100% of GC)
4) Minimum ankle angle (dorsiflexion) Posterior-Anterior axis (0-20% of GC)
5) Positive impulse GRF Posterior-Anterior axis (0-100% of GC)
6) Negative impulse GRF Posterior-Anterior axis (0-100% of GC)
7) Maximum peak GRF Distal-Proximal axis (0-100% of GC)
8) Slope of the GRF Distal-Proximal axis (0-11% of GC)

No differences could be detected in the remaining variables (see table 3.2), suggesting the performance of the feet based on those variables was comparable. The small sample size (N = 5) of this study should be considered when interpreting the results.

5.7 Conclusions

No large differences were noted in the gait pattern of the either foot. When all biomechanical variables were considered the SAFE prosthetic foot had some biomechanical advantages over the NPO. These included reduced vertical GRF peak as well as larger, smoother braking and propulsion impulses. The SAFE foot produced patterns closer to those of able-bodied individuals and show less variation between the residual and sound limbs. Many researchers feel that the performance of a foot prosthesis can be measured by how closely it mimics the able-bodied gait patterns. The inability to detect differences in 32 of the variables suggests that the NPO foot's performance did not differ from the SAFE foot for these variables.

Although results from the gait testing suggest that the NPO foot is slightly less efficient and possibly a less stable prosthesis, no deviation was large enough to cause the researcher to conclude that the gait pattern produced by the NPO foot would be detrimental to the user. The NPO foot did have the advantages of creating a higher degree of dorsiflexion and produced lower knee joint moments than the SAFE foot. The overall performance of the NPO foot was acceptable.

Other factors beside biomechanical measures must be considered when evaluating the acceptability of the NPO prosthetic foot. Unlike the SAFE foot, the NPO foot was specifically designed for use in 3rd world counties. Research and design in the field of prosthetics is an industry largely driven by the interests of developed countries. Design and research therefore consider the needs and prosperity of these developed countries. Research aims to "optimize" the gait produced by prosthetics. The general trend in prosthetics has been towards more intricate multi-piece designs that use sophisticated energy return materials (Rubin, Ficher, et al. 1986). The result of this trend has been large increases in the cost of modern prosthetics due to material cost, production cost, and the requirement for skilled individuals to fit and maintain these prosthetics. New specialized custom build prosthetics designed to allow amputees to participate in sports can cost in excess of five thousand dollars. The cost of normal use prosthetic feet is also

on the rise. Prices for many of the most commonly used feet range from $150 - $800 dollars. Although the impact of this trend is not as dramatic in countries with developed health care systems and/or health insurance, the same can not be said of developing countries that lack such amenities.

The need to consider issues other than gait performance must be addressed when designing a prosthetic foot for the amputee population in developing countries (Sethi 1989)(Cummings 1996). For a prosthetic foot to meet the needs of a developing country it should be low cost, durable, simple to repair, uncomplicated and appropriate for the specific climate and culture of the country in which it is introduced (Poonekar 1992). The concept behind the NPO foot was to produce a foot with adequate gait properties that would address these needs.

The country of El Salvador has been selected as a possible location for field testing of the NPO prosthetic foot. The needs of the amputee population in El Salvador are common to other developing countries. Large numbers of unexploded landmines remain throughout the country, following a civil conflict in the 1980's. Because of this the number of amputation performed in El Salvador each year is proportionally higher than it is in North America (United States Bureau of Foreign Aid 1989) (Davies, Friz, et al. 1970).

The United States Foreign Aid Service reports that the per capita income of many Central and South American countries, adjusted to reflect the cost of living, place them at lowest in the hemisphere, well below levels of North America. This means that even prosthetics which are considered inexpensive by North American standards are unaffordable to the majority of amputees in less developed countries. As a result, individuals are walking on the pylon only (peg), or crudely fashioning wood feet because they cannot afford to replace the components after the original prosthetic foot has worn out. The number of professionals trained in the fitting and repair of prosthetic devices falls short of the need. Complicated mechanical designs that require professional attention for adjustment are less suitable in these countries (Sethi 1989). This combination of increased numbers of amputees, lower annual incomes and a shortage of individuals skilled in prosthetics in Central American countries such as El Salvador creates the need for a prosthetic foot designed with these factors in mind.

The NPO foot's single piece design and its production from a material that is injection moldable gives the NPO foot with several advantages over the SAFE when considering its uses in 3rd world countries. The projected cost of the NPO foot is between $7 and $10, as compared to the cost of a SAFE prosthetic foot at approximately $70-$80, a substantial cost saving. There are several reasons for the reduced projected cost of the NPO prosthetic foot. The material proposed for the NPO foot (polyethylene) is relatively inexpensive and can be purchased in bulk. The feet will be produced using a process known as injection molding. Polyethylene is melted and injected into a shaped mold to produce the prosthetic foot. This is only possible because the prosthetic foot is a single piece design. Unlike the SAFE prosthetic foot, the production of the NPO foot is a one step process. The heel of the SAFE prosthetic foot is made from a different material than the keel. These two parts must be jointed together after they are produced independently adding processing steps and therefore increasing production costs. A cosmetic covering is added to the SAFE prosthetic foot. Although this may enhance the esthetic appearance of the foot it also increases costs.

The single piece design offers several other advantages. The NPO foot is expected to be more durable and have a longer service life than the SAFE prosthetic foot. Many prosthetic feet experience mechanical failure at the attachment points between parts. The single piece design means that there are no attachment points between pieces that can be areas of deterioration. Although the hard surface of the NPO prosthetic foot has been shown to be a less effective shock absorber, it will be more resilient to wear. When the SAFE prosthetic foot is worn outdoors, without a shoe, the spongy heel is vulnerable to high wear. The simplicity of the NPO design makes this foot uncomplicated which addresses the concern of a shortage of skilled individuals in underdeveloped countries to fit and repair prosthetics.

Studies conducted for other prosthetic feet designed for use in underdeveloped countries found that the acceptance of the foot can be affected by socio-cultural concerns. The two most prominent are culturally specific functional requirements and cultural acceptance of appearance (Meanley 1995)(Cummings 1996)(Bartkus, Colvin, et al. 1994).

An example of culturally specific functional requirements is found in India. The ability to sit cross legged has been identified as an important aspect of Indian culture. Prosthetic feet that do not allow rotation about the distal-proximal axis prevent this activity and are rejected in such cultures regardless of gait performance. The Jaipur foot, introduced into the India market, is an example of the foot design whose success can be attributed to its ability meet culturally specific needs (Arya, Lees, et al. 1995).

A common function replaced by a prosthetic foot is one of esthetics. The degree to which this property is important differs between cultures but is always present. Cultures in underdeveloped countries often value esthetics (ability to pass for real) more highly than North American culture. This can be more important than the gait performance of a foot in some cultures. The cosmetic cover of the SAFE prosthetic foot enhances its appearance by making the shape and color of the prosthetic foot more closely resemble that of a real foot. There are no plans to add a cosmetic cover to the NPO foot due the large increase in cost that this would create. As a result the NPO prosthetic foot will not as closely resemble a real foot. This fact could adversely affect the NPO's acceptance in certain cultures.

Although the biomechanical performance of the NPO foot was found to be less than the SAFE foot in some respects its overall biomechanical performance was deemed acceptable. When non-biomechanical factors are considered the acceptable gait mechanics combined with low cost, high endurance and simplicity make the NPO prosthetic foot an excellent choice for underdeveloped countries. Whether or not the NPO can achieve acceptance in these cultures remains to be seen. With some design changes to improve the stability of the prosthetic foot at heel strike acceptance into North American and European markets may also be possible.

Appendix A

Pilot Trial Report

A pilot study was conducted prior to the collection of test data. The purpose of this pilot test was twofold; firstly to ensure that the test protocol was sound, and secondly to determine if any last minute design alteration of the NPO prosthetic foot were required. Pilot testing was performed on December 8, 1998 with aid of a BK amputee subject. Two testing sessions were performed, one for the SAFE prosthetic foot, and one for the NPO prosthetic foot. The subject performed seven trials at each testing session. The best five trial of seven were chosen based on visual observation of the data.

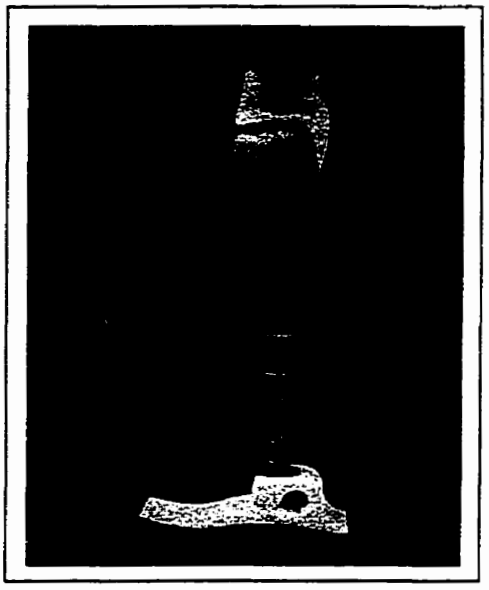
When the gait data from the two prosthetic feet was compared, it was found that the NPO foot produced a smaller fore/aft deceleration impulse, and exhibited a different vertical ground reaction force pattern than the SAFE foot. The subject reported that he found the NPO foot felt somewhat stiff, and that heel strike seemed to occur prematurely. This information was provided to the designers of the NPO prosthetic foot (Niagara Prosthetics and Orthotics) in the form of an "Interim Report".

The NPO design underwent slight modification based on the information provided by both the empirical gait data and the subjective feedback from the subject.

The testing protocol was found to be acceptable and no changes were implemented.

Analysis of a Low Cost Prosthetic for use in Developing Countries

NPO Prosthetic Foot Design



Interim Report

December 8,1998

SUBJECT

Subject chosen was found to meet all the following:

- Male
- Early 30's
- Active and mobile
- Uses several different high function foot prosthetics: -Flex Foot
  -Carbon Copy II
  -Endolite
  -SACH
  -SAFE

- No major stump abnormalities
- No major problems with the contralateral leg and/or joints of contralateral leg
- No major gait abnormalities outside of amputation

Testing Protocol

Gait Analysis:

A certified prosthetist fitted and adjusted the NPO prosthetic foot for the subject. The subject's existing socket was used to ensure proper fit and function for the trial. Ground reaction force data for gait was collected using an AMTI force plate. Segmental and joint movement data were collected using optotrack (optoelectric motion tracking system). A series of 10 marker locations were selected and fitted with infrared emitting diodes. Landmarks used were those found in literature to be commonly used in the study of both normal and amputee gait. Most marker locations were bony landmark sites. Location of these landmarks on the residual limb and prosthetic were estimated, using the sound limb as reference.

Marker Locations:

1. Hip (Greater trocanter)
2. Thigh (Mid quadriceps)
3. Superior Knee (Lateral condyle of femur)
4. Inferior Knee (Lateral fibular head)
5. Shank (Mid tibia)
6. Malleolus
7. Calcaneus
8. Instep (Mid longitudinal arch)
9. 5th metatarsal
10. 5th toe (Lateral)

Collection Parameters:

Optotrack	Force Plate:
Collection rate – 100 Hz	Collection rate – 100 Hz
Collection time – 4 Sec	Collection rate - 100 Hz
	Gain - 2000

Subjective Reports

During testing the subjects was asked to give feedback on the foot with regards to comfort, ease of transition, and overall performance. The subject commented that the overall performance of the foot was good. He found the transition between feet relatively easy and experienced little difficulty in adapting to gait on the new foot. The subject commented that the mechanics of the foot did not seem overly dissimilar from those of other prosthetics that he has used. When asked what improvements could be made to the foot, the subject reported that he found the foot "a little too stiff". The subject also commented that although he experienced no difficulties while traveling up stairs, travelling down was somewhat awkward. The subject also commented that because the heel of the NPO foot was square rather than rounded, he experienced some "kicking". If the subject heel struck with a corner rather than squarely on the heel, an internal or external rotation of the foot was seen.

During testing the subjected exhibited gait that appeared smooth and natural, with no major hesitations. No noticeable cadence differences were observed while using either prosthetic.

Results

The results are separated into four sections:
1. Marker movement (optotrack)
2. Center of pressure of prosthetic (force plate)
3. Ground reaction force data (force plate)
4. Ankle angle data (optotrack)

Marker Movement:

All optotrack motion data was normalized to 100 data points. Data from the first six markers was separated into fore/aft (X), Medio-Lateral (Y), and Vertical location (Z). Four trials each for the NPO and Safe feet were graphed to compare the affect of each prosthetic on movement patterns of the markers. Each plot represents one full gait cycle from heel strike to heel strike. Data is presented as % of gait cycle (100% or 100 data points).

Patterns of movement for all six markers were found to be very similar. Movement patterns in fore/aft motion, and vertical motion were almost identical. Some small variations can be seen in medio-lateral movement of some markers.

Center of Pressure:

Center of pressure patterns for both feet appear similar with the NPO foot exhibiting less mediolateral movement. Because the NPO foot is flat and uniform along its bottom surface, mediolateral rolling seen in normal gait is not produced. The safe foot is contoured to simulate a longitudinal arch of a normal foot, and therefore may allow more mediolateral roll.

Ground Reaction Force Data:

Ground reaction force data acquired from the force plate was normalized to 60 points. Because the force plate is only able to measure forces applied to it while the foot is in physical contact with it, only stance phase data is collected. Stance phase represents ~60% of the gait cycle. Normalization to 60 points allowed data to be plotted as % of gait cycle. Data is separated and graphed as x, y, z, or fore/aft, medio-lateral, and vertical force.

In all three planes raw data for the NPO foot can been seen to exhibit increase high frequency noise when compared to the Safe foot. It is most likely that this is due to increased impact vibration, due to reduced cushioning. The Safe foot is design to provide cushioning upon impact with a softened heel, the NPO has no such feature. The subject also wore an athletic shoe over the Safe foot, providing further impact cushioning.

Medio-Lateral : A small increase in medio-lateral GRF can be seen when the averages of the safe and NPO foot are compared.

Fore/Aft : The graph of the safe foot fore/aft GRF appears very similar to the GRF of normal gait. A smooth change of sign (from + to -) can be seen representing braking and propulsive forces. A notable decrease in braking force can be seen in the NPO fore/aft GRF. Propulsion patterns for the two feet appear similar.

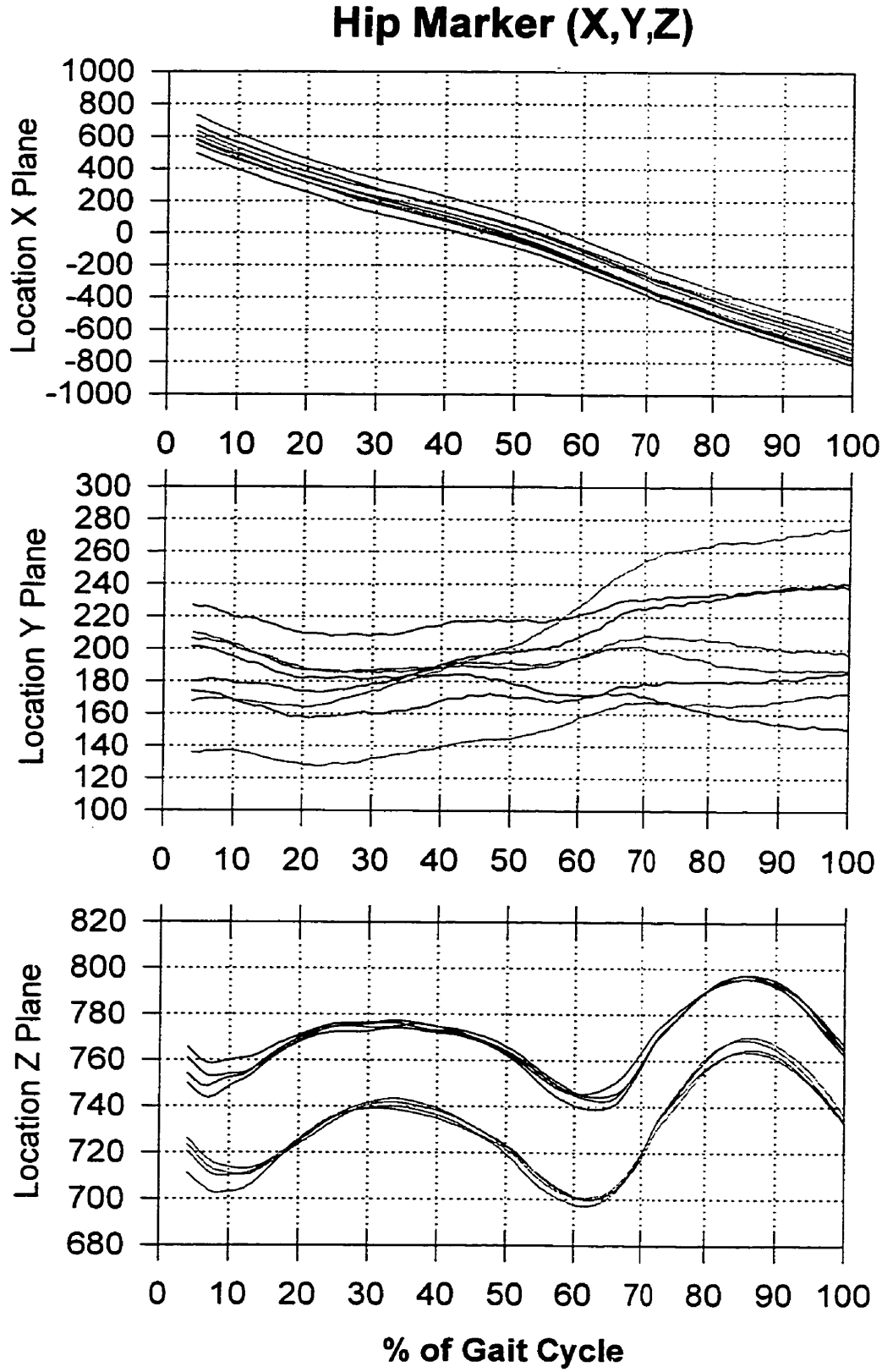
Vertical: The vertical pattern of the safe foot closely mimics that of normal gait. A clear smooth curve with two distinct peaks representing heel strike and toe off can be seen. There is a clear distinction between these peaks and the valley representing midstance. Although a similar pattern can be seen in the NPO foot it is much less pronounced. The difference between the peaks and midstance is much smaller.

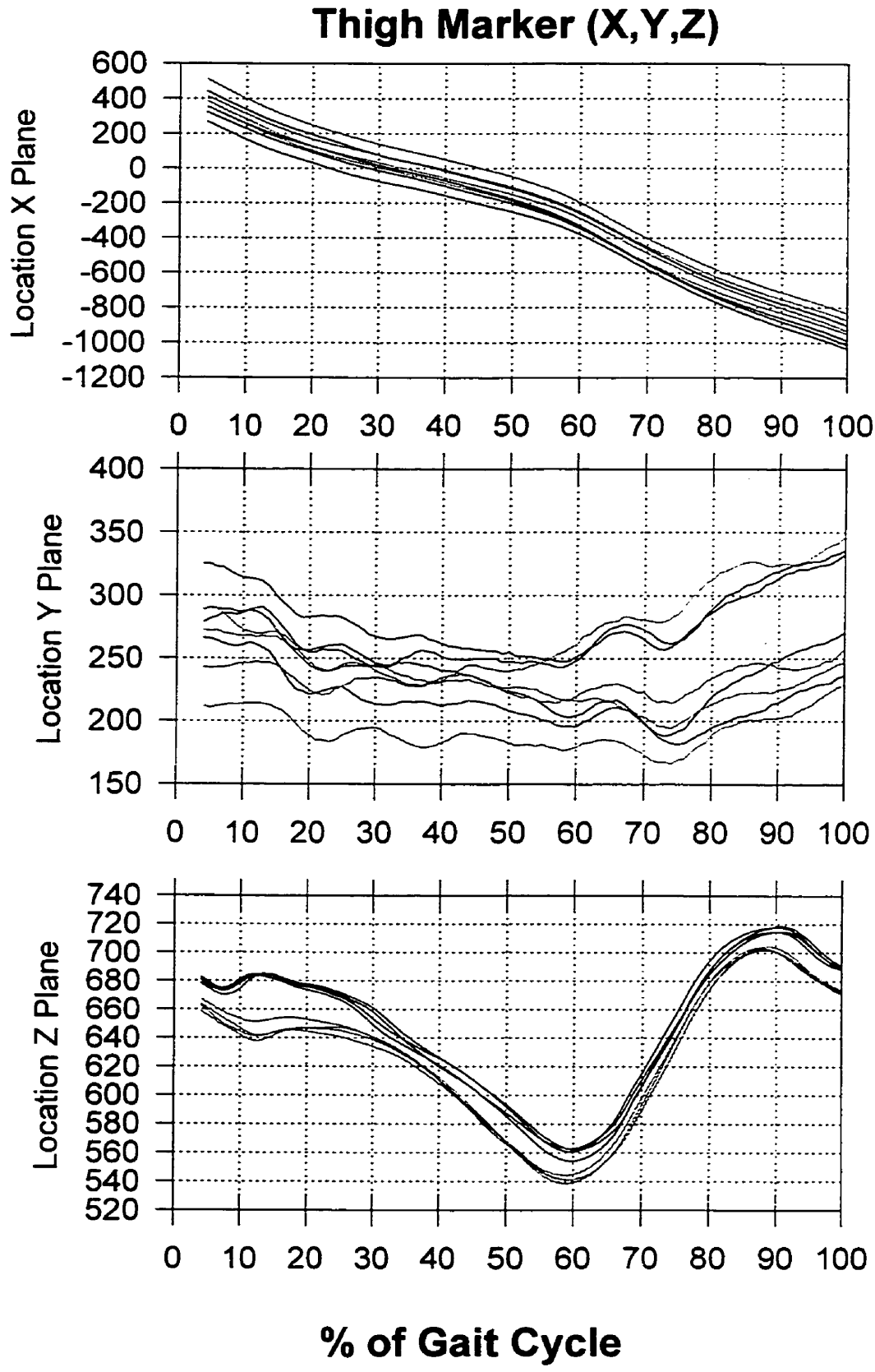
Ankle Angle

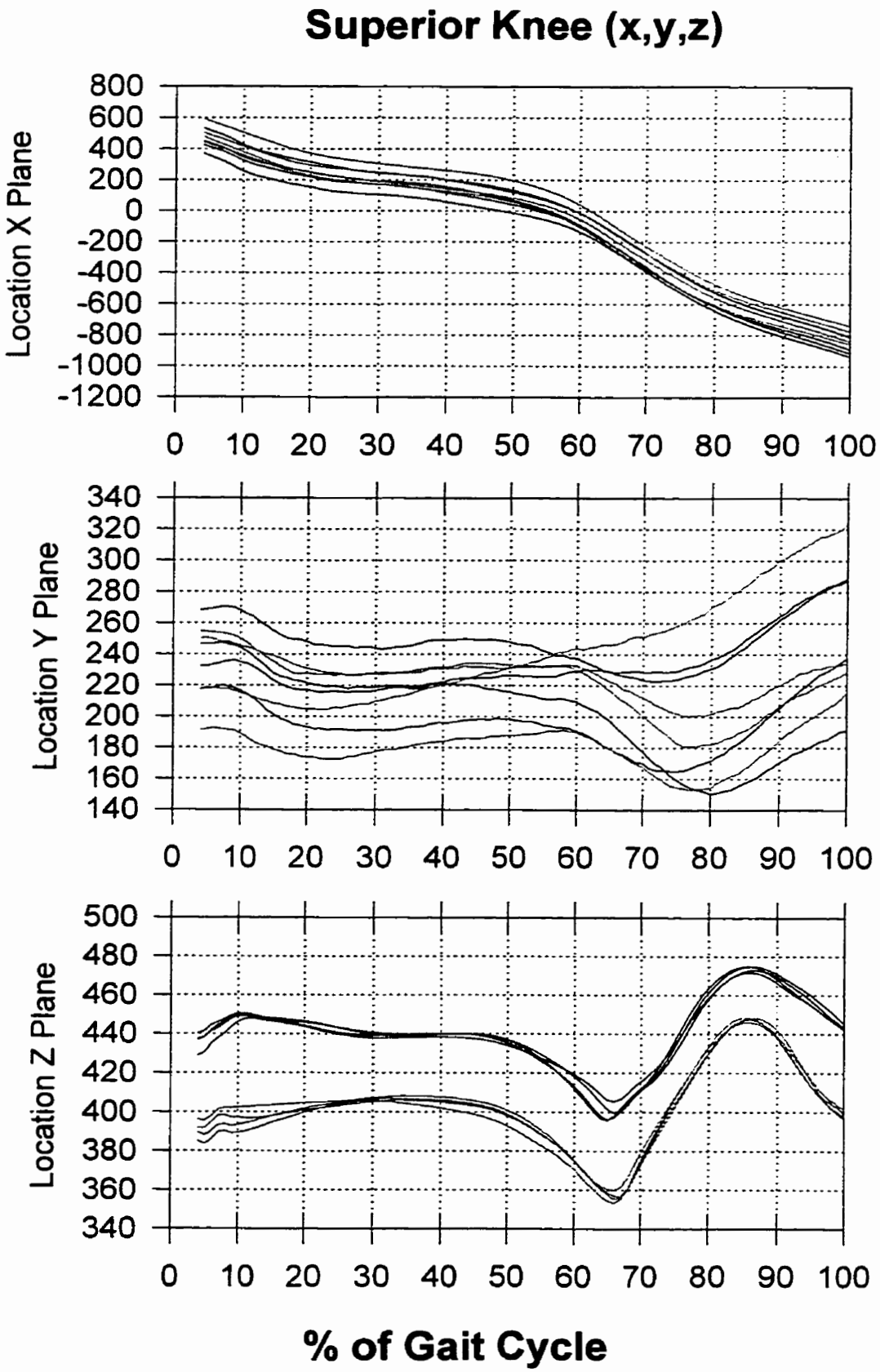
Ankle angle data was normalized to 100 points to represent one complete gait cycle. Data is graphed from heel strike to heel strike. Data is compared to normal data for ankle angles (Winter, 1979).

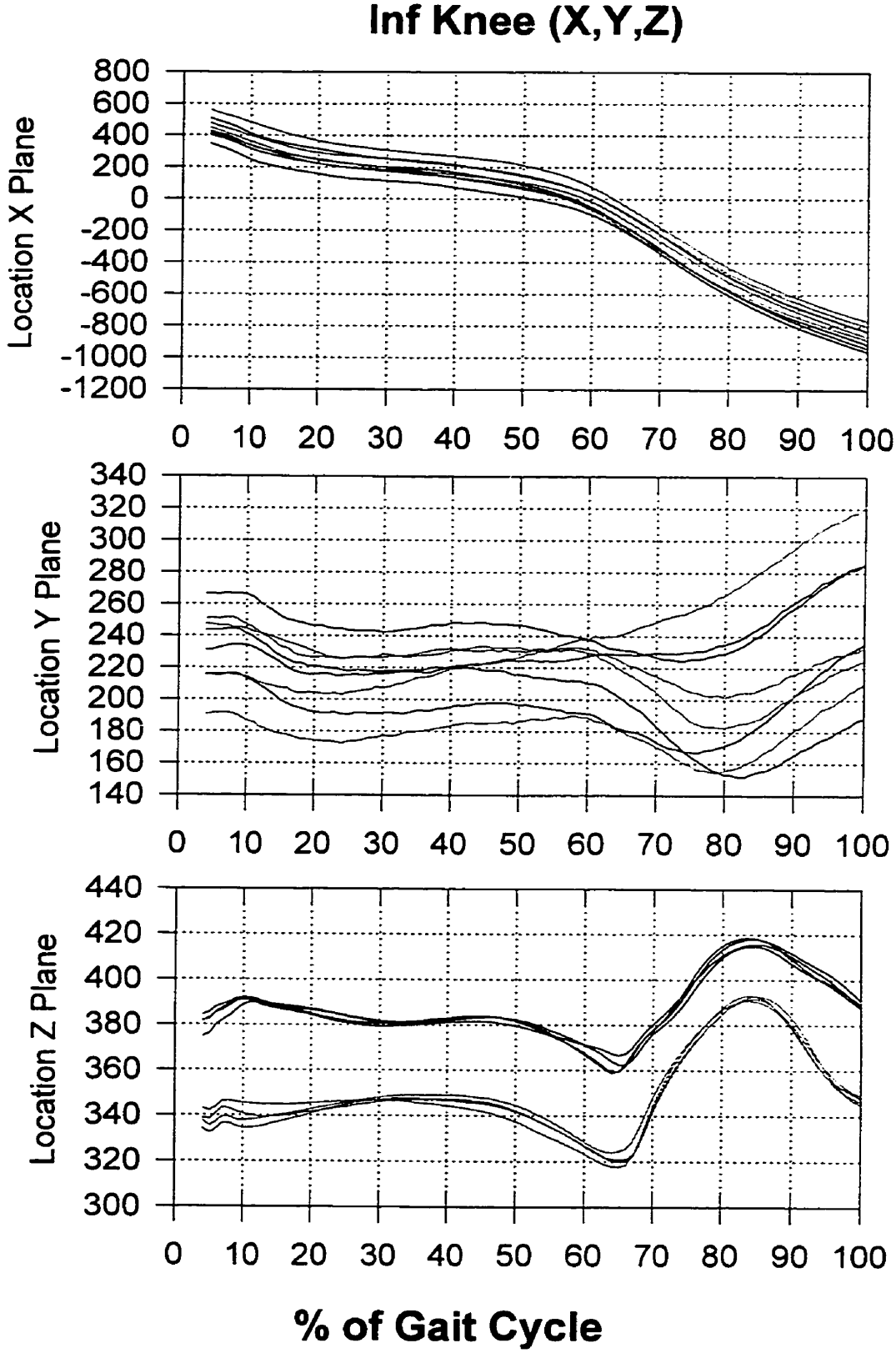
Ankle angles from heel strike to flat foot appear to follow patterns similar to normal gait. There is a large difference in ankle at toe off. This is due to the fact that the amputee can not use calve muscle to produce dorsiflexion in preparation of swing phase.

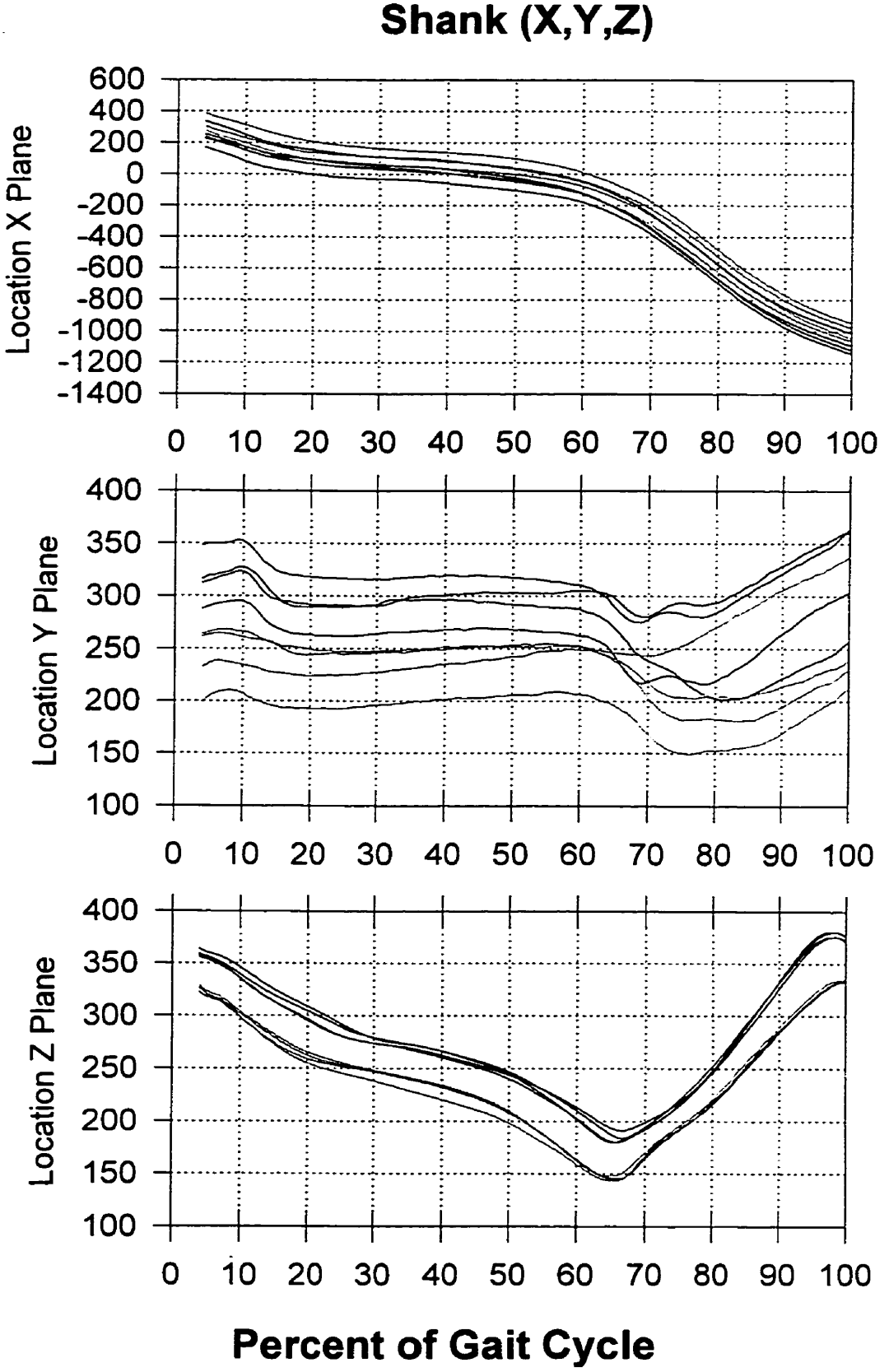
Optotrack Data
Marker Motion Patterns

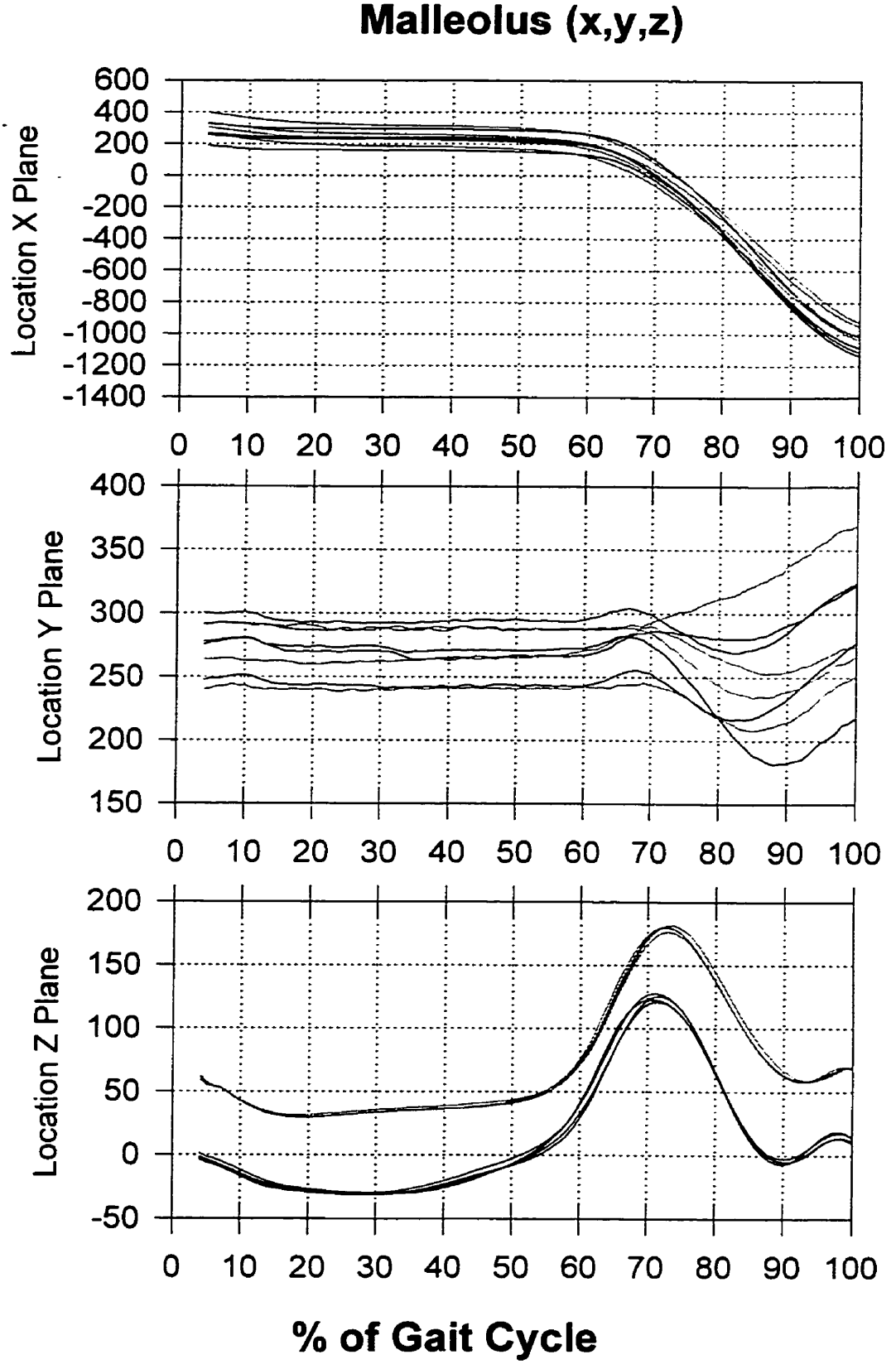

Hip Marker (X,Y,Z)
Location X Plane
1000
800
600
400
200
0
-200
-400
-600
-800
-1000
0 10 20 30 40 50 60 70 80 90 100
Location Y Plane
300
280
260
240
220
200
180
160
140
120
100
0 10 20 30 40 50 60 70 80 90 100
Location Z Plane
820
800
780
760
740
720
700
680
0 10 20 30 40 50 60 70 80 90 100
% of Gait Cycle


Thigh Marker (X,Y,Z)
Location X Plane
600
400
200
0
-200
-400
-600
-800
-1000
-1200
0 10 20 30 40 50 60 70 80 90 100
Location Y Plane
400
350
300
250
200
150
0 10 20 30 40 50 60 70 80 90 100
Location Z Plane
740
720
700
680
660
640
620
600
580
560
540
520
0 10 20 30 40 50 60 70 80 90 100
% of Gait Cycle

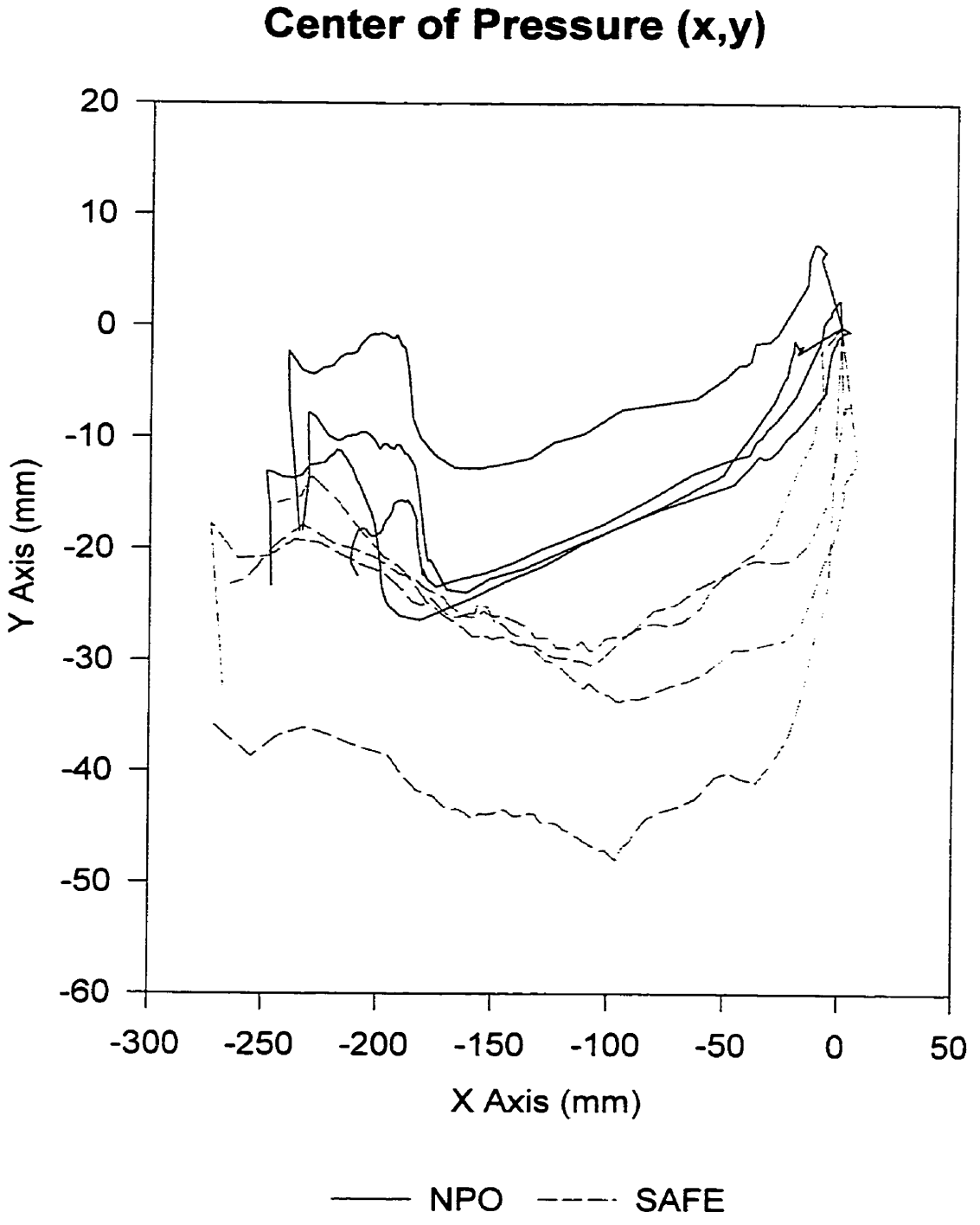

Superior Knee (x,y,z)
Location X Plane
800
600
400
200
0
-200
-400
-600
-800
-1000
-1200
0 10 20 30 40 50 60 70 80 90 100
Location Y Plane
340
320
300
280
260
240
220
200
180
160
140
0 10 20 30 40 50 60 70 80 90 100
Location Z Plane
500
480
460
440
420
400
380
360
340
0 10 20 30 40 50 60 70 80 90 100
% of Gait Cycle


Inf Knee (X,Y,Z)
Location X Plane
800
600
400
200
0
-200
-400
-600
-800
-1000
-1200
0 10 20 30 40 50 60 70 80 90 100
Location Y Plane
340
320
300
280
260
240
220
200
180
160
140
0 10 20 30 40 50 60 70 80 90 100
Location Z Plane
440
420
400
380
360
340
320
300
0 10 20 30 40 50 60 70 80 90 100
% of Gait Cycle

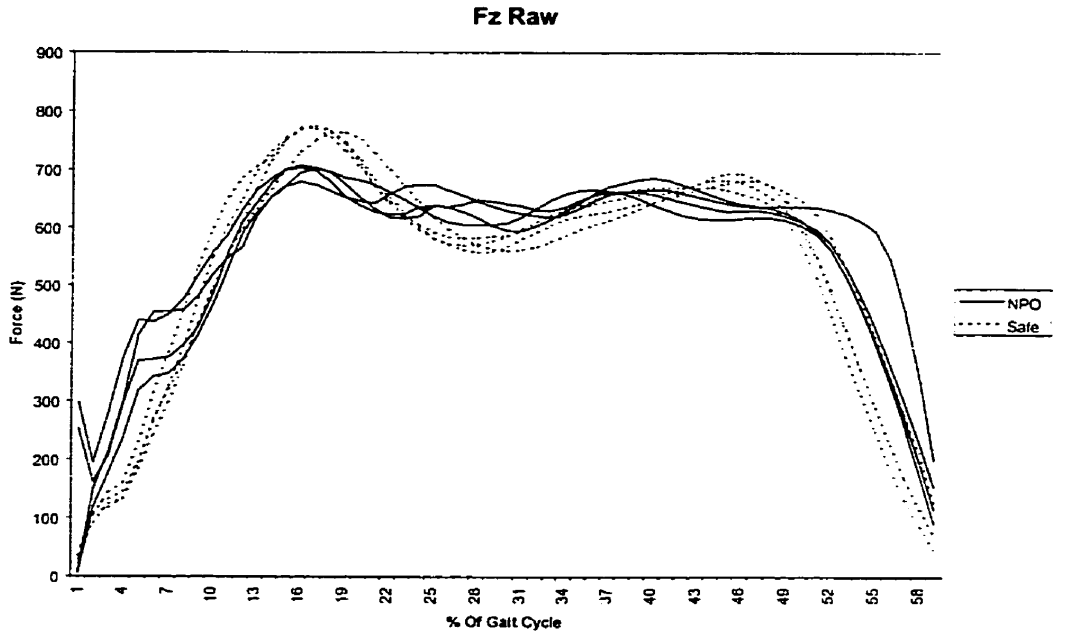

Shank (X,Y,Z)
Location X Plane
600
400
200
0
-200
-400
-600
-800
-1000
-1200
-1400
0 10 20 30 40 50 60 70 80 90 100
Location Y Plane
400
350
300
250
200
150
100
0 10 20 30 40 50 60 70 80 90 100
Location Z Plane
400
350
300
250
200
150
100
0 10 20 30 40 50 60 70 80 90 100
Percent of Gait Cycle

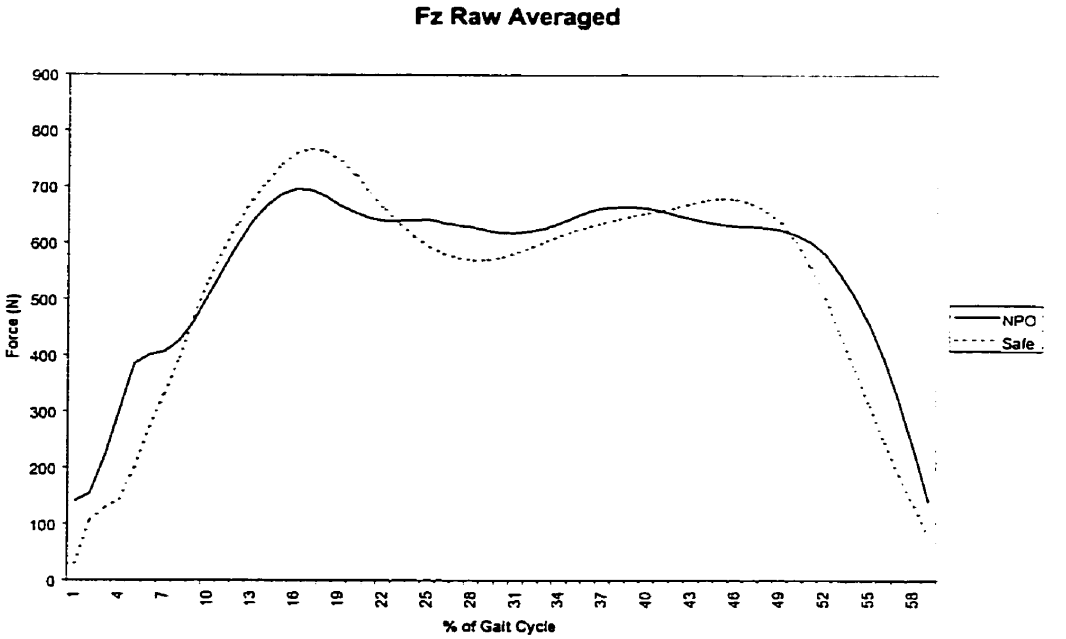

Malleolus (x,y,z)
Location X Plane
600
400
200
0
-200
-400
-600
-800
-1000
-1200
-1400
0 10 20 30 40 50 60 70 80 90 100
Location Y Plane
400
350
300
250
200
150
0 10 20 30 40 50 60 70 80 90 100
Location Z Plane
200
150
100
50
0
-50
0 10 20 30 40 50 60 70 80 90 100
% of Gait Cycle

Force Plate
Center of Pressure

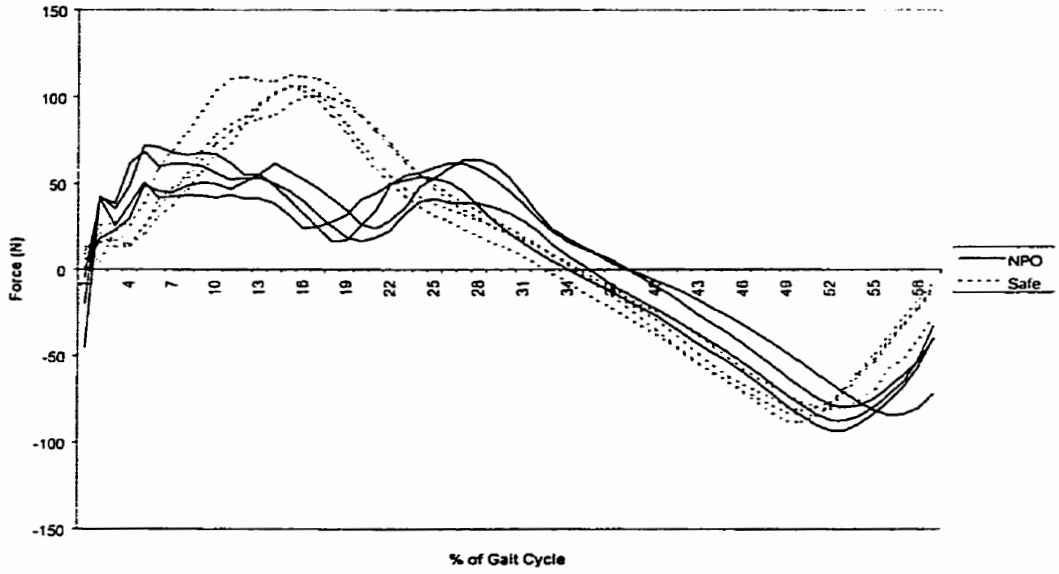

Center of Pressure (x,y)
20
10
0
-10
-20
-30
-40
-50
-60
Y Axis (mm)
-300
-250
-200
-150
-100
-50
0
50
X Axis (mm)
NPO
SAFE

Force Plate
Ground Reaction Forces

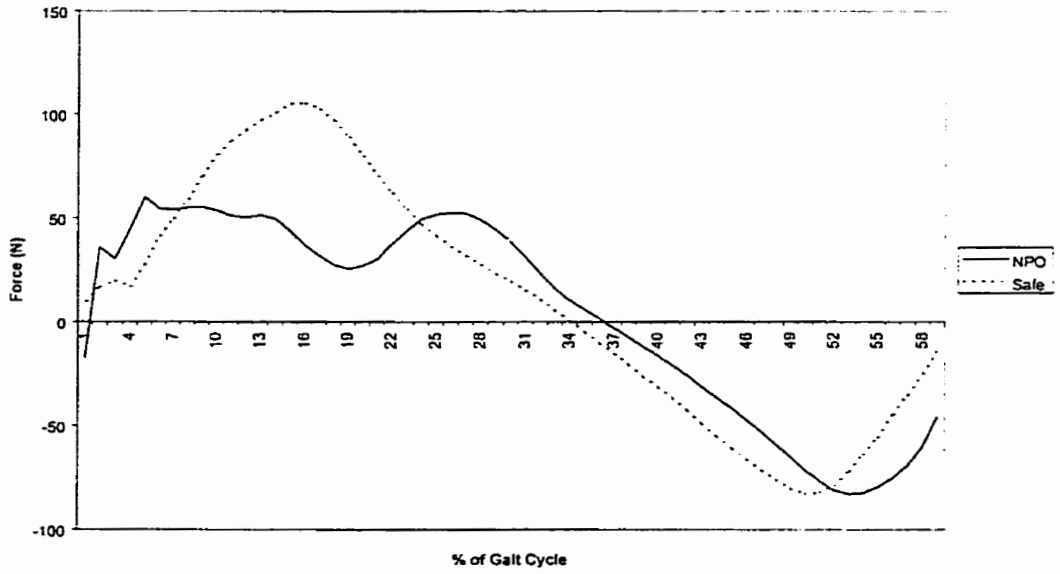

Fz Raw
900
800
700
600
500
400
300
200
100
0
Force (N)
1
4
7
10
13
16
19
22
25
28
31
34
37
40
43
46
49
52
55
58
% Of Gait Cycle
NPO
Safe


Fz Raw Averaged
900
800
700
600
500
400
300
200
100
0
Force (N)
1
4
7
10
13
16
19
22
25
28
31
34
37
40
43
46
49
52
55
58
% of Gait Cycle
NPO
Safe

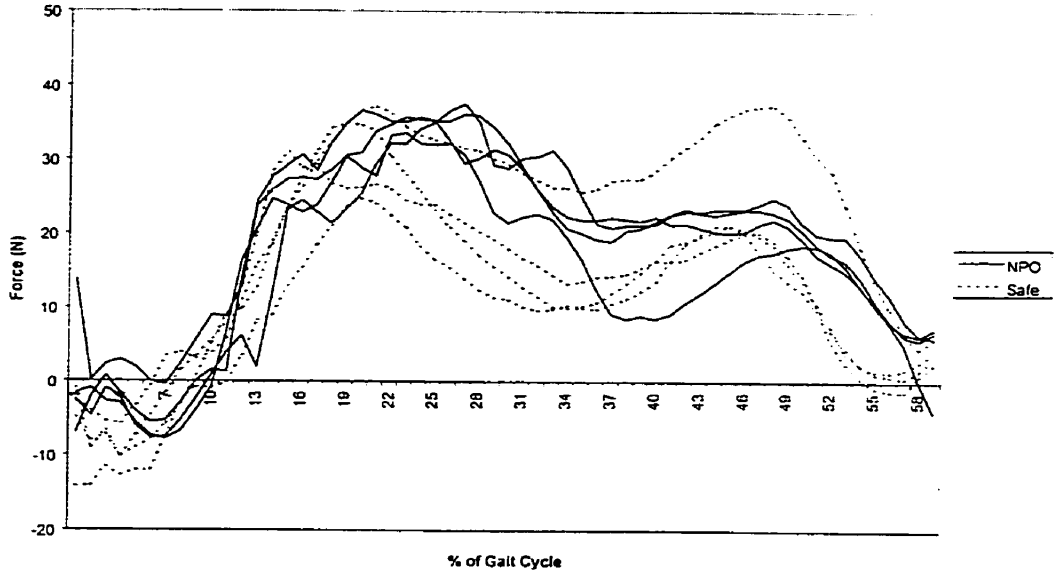
Fx Raw



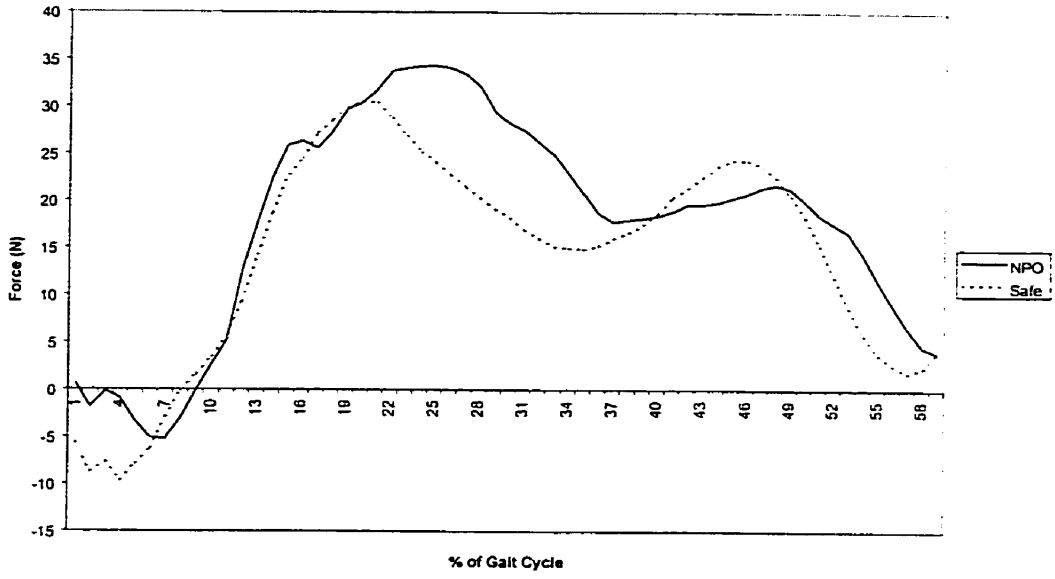
Fx Raw Averaged



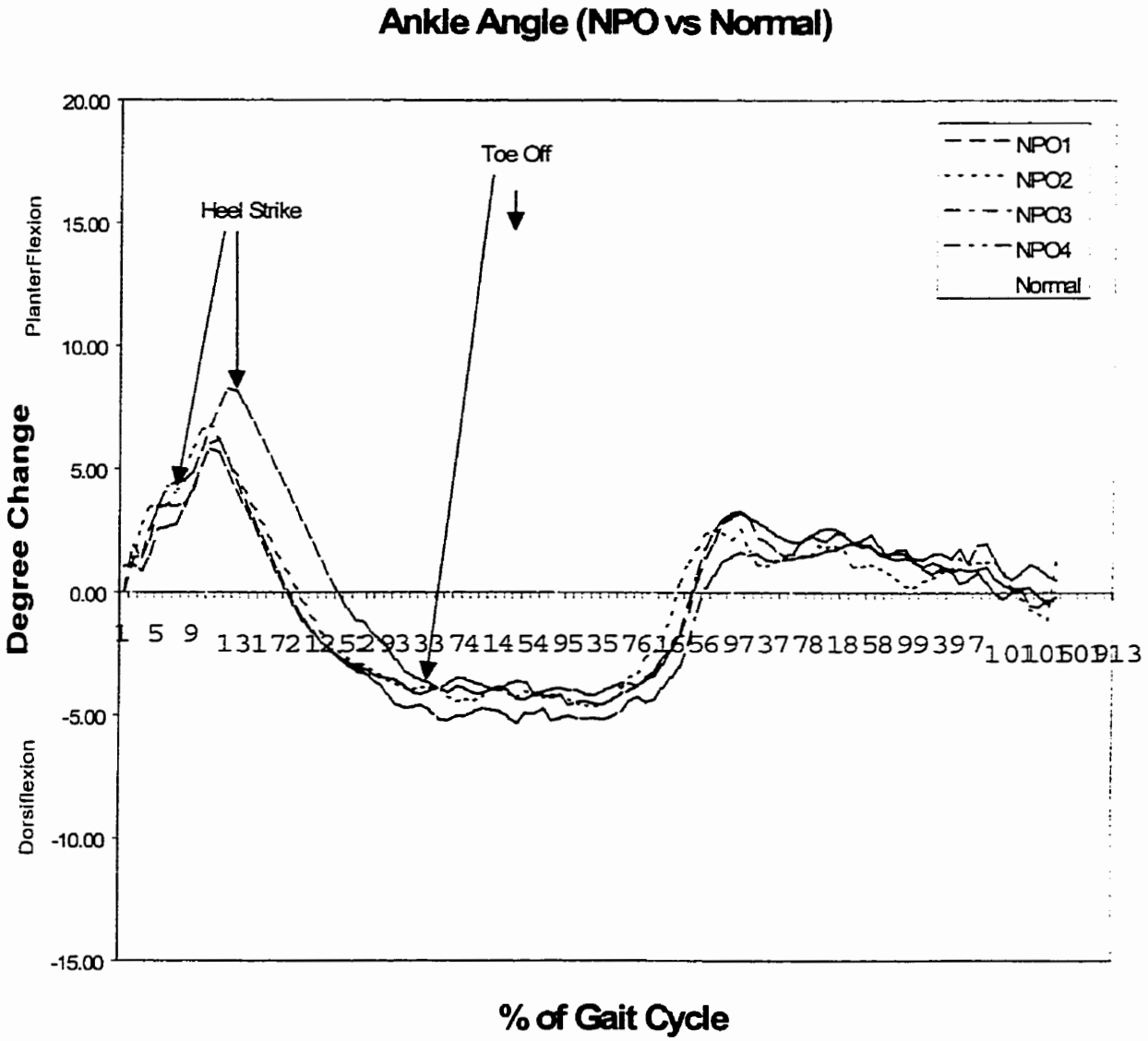
Fy Raw



Fy Raw Averaged



Optotrack
Ankle Angle Data


Ankle Angle (NPO vs Normal)
20.00
15.00
10.00
5.00
0.00
-5.00
-10.00
-15.00
PlanterFlexion
Dorsiflexion
Degree Change
Heel Strike
Toe Off
NPO1
NPO2
NPO3
NPO4
Normal
1 5 9 13 17 21 25 29 33 37 41 45 49 53 57 61 65 69 73 77 81 85 89 93 97 101 105 109 113
% of Gait Cycle

Appendix B

Consent Form

Consent Form Code:_____

Evaluation of a Low Cost Foot Prosthetic for Developing Countries

Subjects are invited to participate in a scientific research project designed to evaluate the Niagara Prosthetic and Orthotics (NPO) foot design. Participation in the study is completely voluntary, and subjects are free with draw from testing at any time. Information pertaining to the purpose and protocol of the study are listed below.

Investigators

Derek Potter	531-0369(Home)
	545-2658(School of Phys. Ed.)
Dr. Patrick Costigan	542-2468 (Home)
	545-2666 (School of Phys. Ed.)
Dr. J.T. Bryant	548-2430 (Clinical Mechanics Group)

Study Rationale

The Clinical Mechanics Group at Queen's University is helping to develop and evaluate a new low cost prosthetic foot design. The prosthetic is intended for use in developing countries. Gait analysis will be performed on the new prosthetic foot, and compared to the gait characteristics of subjects with normal gait (non-amputee), and those using existing low cost prosthetic feet. Performance of the new prosthetic will be studied during level walking and while walking and carrying a light load. Gait analysis will be used to compute moments at the knee joint. Clinical test will be used to evaluate functional performance.

Procedures

Subjects will be ask to attend two separate testing sessions. In session 1 subjects will walk on the SACH (Solid Ankle Cushioned Heel) prosthetic. In session 2 the NPO foot prosthetic will be used. Subjects will be provided with feet two days prior to testing and asked to use these feet exclusively. This is to allow the subject to adapt to walking in the new prosthetic. Each testing session will last from 1.5 – 2.5 hours.

Clinical Examination. You will be asked some questions related to your general fitness and some simple anthropometrics, such as height weight, and leg lengths will be measured. You will also be asked to perform a 10 m walking test, and complete a subject questionnaire. Questionnaire will contain questions referring present and prior use of prosthetics, and level of activity.

Gait Assessment. You will be asked to walk on level ground. Small light emitting diodes (LED) will be placed on selected joint landmarks of your leg and a footswitch will be attached to **your** shoe.

Consent Form Code:_____

Statement of risks involved

1. The gait assessment is not harmful and will not cause pain. The amount of walking is not enough to cause fatigue. Load carried during walking will be maintained at or below 15% of body weight. This load is not enough to cause fatigue.

2. Clinical tests including the 10m walking test pose no risk to the subject other than the risks associated with normal walking.

3. All tests will be done in Kingston General Hospital with the attendance of an adequately trained clinician or research assistant.

4. Subjects may experience some pain or discomfort during the two day adaptation period when using and unfamiliar prosthetic.

5. During the adaptation period there may be an increased risk of falling or lose of balance when using an unfamiliar prosthetic.

Maintenance of confidentiality

The identity of the patient is recorded only once by the research assistant at the time of filing the patient consent forms. These files are accessible only to the research assistant and principal investigators. All patients are assigned a record number which is linked to this file. All data recorded in computer files contain this number, rather than the patient name.

In all cases of publication, summary (aggregate) data are used in such a way that no individual can be identified.

All paper data forms will be transferred to electronic data forms and these files will be encrypted. Paper forms will be stored in a locked filing cabinet.

Consent Form Code:_____

Expected Benefits

No direct benefits are expected for the patient or volunteer subjects. Some patients may benefit indirectly as a result of the detailed study of the biomechanics of the new prosthesis. Those who may see direct benefit from study results are the undeveloped countries that may make use of the prosthetic

Consent

I,_____ have reviewed the protocol and rationale for the procedures of gait analysis. I understand what the procedures involve. I understand that I may not benefit directly from the information obtained from this study, but others in need of low cost prosthetics may. I realise that I am free to withdraw from the study at any time, without prejudice or penalty. Should I have concerns about this study I am free to ask any of the research investigators involved, as well as the Head of the Department of Mechanical Engineering, Dr. B. Surgenor (545-2575), the Director of the School of Physical and Health Education, Dr. J. Stevenson (545-2666), and the Chair of the Research Ethics Review Board, Dr. A.Clark, Faculty of Medicine (545-2494).

I acknowledge the receipt of my copy of this form.

Volunteer:_____

Witness:_____ _____
Signature Date

Appendix C

Anthropometric and Correction Vector Data Collection Sheet

Subject Information / Anthropometric

Subject Name_____ Date_____

Height:_____

Weight_____

Floor to Greater Trochanter_____

Floor to Tibial Plateau _____

Upper Thigh Circumference _____

Circumference of Calf _____

BONE WIDTHS

Femoral (condyle to condyle) _____

Ankle _____

Reference Position Angle _____

Side Flexion Angle (Lateral) _____

(Frontal) _____

Foot Rotation _____

CORRECTION VECTORS

	Fibula	Knee	Hip
X			
Y			
Z			

Appendix D

Subjective Questionnaire Form

Date_____

Subject Questionnaire

Name_____ Subject Code_____

Height_____

Weight_____

Leg Amputated Left ☐ Right ☐

Cause of Amputation_____

Type of Prosthetic(s) currently used [please list in order of most used – least]

1 _____

2 _____

3 _____

4 _____

Activity Level

Sedentary ☐

Somewhat active ☐

Active ☐

Very Active ☐

Athletic ☐

Elite Athlete ☐

Evaluation of Prosthetic SACH ☐ NPO ☐

Comfort of the prosthetic

Excellent Average Poor

10 9 8 7 6 5 4 3 2 1

Ease of Use

Excellent Average Poor

10 9 8 7 6 5 4 3 2 1

Ease of Adaptation

Excellent Average Poor

10 9 8 7 6 5 4 3 2 1

Stability when Standing

Excellent Average Poor

10 9 8 7 6 5 4 3 2 1

Stability when Walking

Excellent Average Poor

10 9 8 7 6 5 4 3 2 1

Minimizes muscular effort

Excellent Average Poor

10 9 8 7 6 5 4 3 2 1

Heel Strike Feels Good

Excellent Average Poor

10 9 8 7 6 5 4 3 2 1

Toe Off Feels Good

Excellent Average Poor

10 9 8 7 6 5 4 3 2 1

Opposite Leg Feels Good

Excellent Average Poor

10 9 8 7 6 5 4 3 2 1

Limb/Socket Contact is Good

Excellent Average Poor

10 9 8 7 6 5 4 3 2 1

Do you feel you could use this foot to walk on for an entire day Y ☐ N ☐

Comments:_____

_____

_____

_____

_____

_____

Appendix E

Calculation of Mass Moment of Inertia for A Prosthetic Limb

The QGAIT program uses regression equations that use standardized anthropometrics to calculate the following parameters:

1) The mass of the shank
2) The location of the shank center of mass
3) The mass moment of inertia of the shank

The equations based on cadaver studies, estimate the parameters required to calculate knee forces and moments. Prosthetic limbs differ from normal limbs in density, size and shape, and therefore, the regression equations would estimate incorrect values for the three parameters listed above. To account for this difference the QGAIT program was modified (by Dr. Pat Costigan) to allow these parameters to be entered by the user, rather than calculated by the program. Mass and the location of center of mass from each prosthetic was measured for each subject. The mass moment of inertia was not measured; instead a reasonable estimate was made that was entered into the program.

A pendulum frequency test was used to calculate moments of inertia from the test limbs. The prosthetic feet used in the main experiment were used in this experiment. While three prosthetic foot sizes were used in the main study (26 inches, 27 inches, and 29 inches), only the 27 inch NPO and SAFE feet were used in the inertia sub-study. However, to improve the inertia estimate, couplings and titanium/steel pylons of the same length, size, and configuration as those of the subjects were used. Sockets owned by the subjects were not available for testing. Only one standard socket only was available for

testing. This socket was attached to each pylon – foot combination creating six prosthetic limbs for testing.

Methods

Two prosthetic feet were used, a 27 inch NPO and a 27 inch SAFE. Pylon length was measured from the top of the pylon where it connects to the coupling to the bottom of the foot. Three pylon lengths were chosen for testing. These lengths corresponded to 1) the pylon length of the shortest subject, 2) the average pylon length of all 5 subjects, and 3) the pylon length of the tallest subject. These lengths used were:

1) 422 mm
2) 450 mm
3) 505 mm

A knife edge was used to locate the center of mass along the long axis (Distal - Proximal) of the prosthetic limb. The joint line, or the level at which rotational flexion occurs in the prosthetic was estimated, and a line was drawn on the socket. The socket was then attached to the pendulum at this point. Two IREDs (Infra-Red Emitting Diodes) were attached to the prosthetic limb. IRED 1 was affixed at the joint line, where the pendulum attached to the prosthetic limb. IRED 2 was affixed to the bottom of the foot (Figure E1).

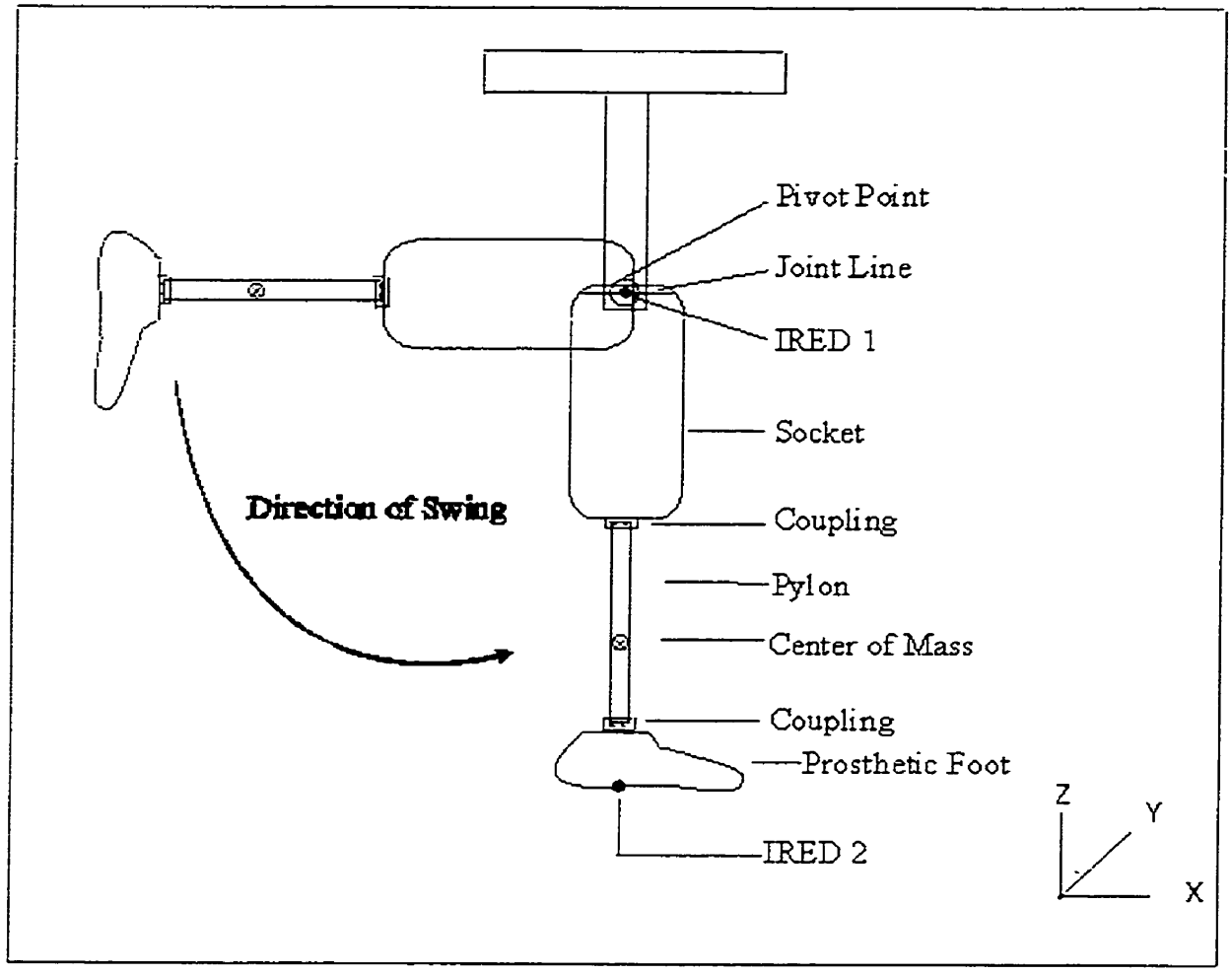


Figure E1: Prosthetic limb pendulum setup

The prosthetic limb was raised to a point so that the pylon was parallel to the ground. It was then released and allowed to swing freely. The Optotrack collected data from IRED 1 and 2 at 100hz for a period of eight seconds. Five trials were repeated for each of the six prosthetic limb configurations.

Information of the location of IRED 2 in the posterior-anterior plane (perpendicular to camera) was imputed into a spreadsheet. Periods of the pendulum were located by finding all maximum height points in the data. One period was considered to be from maximum height 1 to maximum height 2 on the same side. An example of this data can be seen in Table E1. Frequency of the period was calculated as follows:

Frequency(s) = (Number of Frames for period) x (1/100 seconds per frame)

Data from the first two seconds was not used to ensure that errors due to improper release of the pendulum were omitted. Periods were calculated using data from three to eight seconds (Figure E2). The frequencies of these within periods were averaged, and were then averaged over the five trials to give a single period frequency value for each of the six prosthetic limbs.

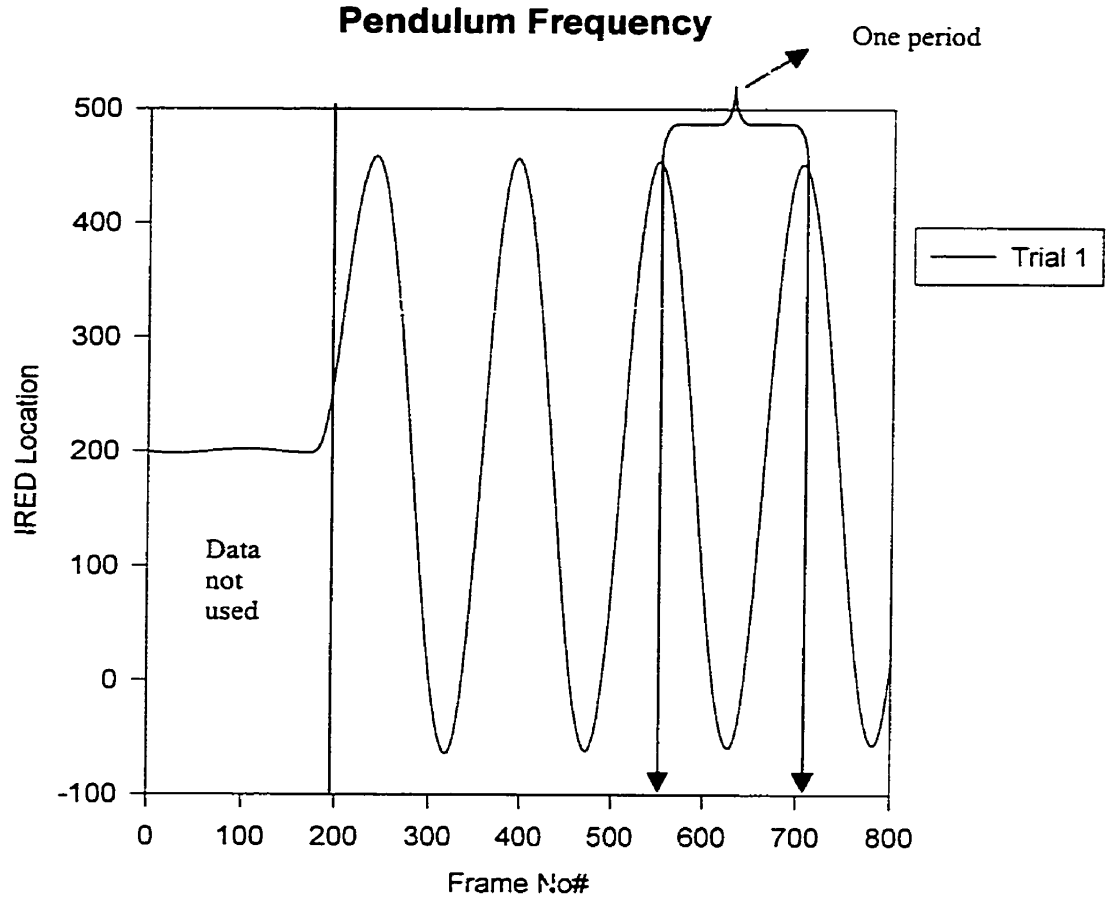


Figure E2: Frequency results from pendulum trials

Mass moment of inertia was calculated using the following equation:

$$I = mag(T/2\pi)$$

Where:

I – Mass moment of inertia (kg/m^2)
m – Mass of prosthetic limb (kg)
a – Distance from pivot point to center of mass of the prosthetic limb (m)
g – Acceleration due to gravity (m/s^2)
T – Frequency of period (s)

Error Estimation

The difference between the SAFE1 and SAFE2 test prosthetic limbs (0.00996 kg/m^2) was greater than the difference between the NPO1 and NPO2 prosthetic limbs (0.00421 kg/m^2). Therefore 0.01kg/m^2 was chosen as a reasonable prediction of error introduced into the data by the use of an estimated value for mass moment of inertia.

Gait data collected for one subject was processed by the QGAIT program using the mass moment of inertia for that subject chosen in this sub-study. Computed values for moments at the knee joint were exported to a spreadsheet. Values for the original data plus predicted error, minus predicted error, plus two times predicted error and minus two times predicted error were calculated and added to the spreadsheet. Formulas in the spreadsheet were used to calculate the Root Mean Squared (RMS) difference between the columns.

Results

Mass moment of inertia for each of the six prosthetic limbs can be seen in Table E1. Mass moments of inertia for the prosthetic limbs with the SAFE foot attached ranged from 0.118024 kg/m^2 to 0.156856 kg/m^2. Mass moments of inertia for the prosthetic limbs with the NPO foot attached ranged from 0.138763 kg/m^2 to 0.207618 kg/m^2.

Table E1: Mass moment of Inertia for test prosthetic limbs

Moment of Inerita (kg/m2)				
SAFE 1	0.122024		NPO 1	0.138763
SAFE 2	0.131984		NPO 2	0.142973
SAFE 3	0.156856		NPO 3	0.207618

Error Estimate

Error was defined as percent error across the full gait cycle. This error was calculated as follows:

% Error = (Root Mean Square (RMS) / Total Range (x, y, z)) x 100%

Maximum possible error was found to be 9.332% in the y (posterior anterior axis), for minus two time the predicted error (Table E2, E2, E3).

Conclusion

Four of the Five subjects who participated had similar prosthetic foot sizes and pylon length measurements. The prosthetic foot size of the subjects CL, MA, MM, and SG were 27 inches, 26 inches, 27 inches, and 27 inches respectively. The pylon to foot measurements for these subjects were 440mm, 431mm, 428mm, and 445mm respectively. The prosthetic test limbs SAFE2 and NPO2 all used the 27 inch foot with a pylon length of 450 mm. It was therefore decided that the mass moment of inertia of these limbs (SAFE2 – 0.131984, NPO – 0.142973) would be the best estimate for subjects CL, MA, MM, and SG.

Subject DK used a 29 inch foot and had a pylon length of 505 mm. The prosthetic test limbs SAFE3 and NPO3 used a 27 inch foot with a pylon length of 505 mm. Mass moment of the inertia calculated from test limbs SAFE3 and NPO3 were chosen as the closest estimation of DK's mass moment of inertia (SAFE3 – 0.156856, NPO3 - 0.207618).

Table E2: Calculation or percent error for moments at the knee joint (PA axis)

X +0.02kg/m	X +0.01kg/m	X Original	X -0.01kg/m	X -0.02kg/m
$(X_1 - X_3)$	$(X_2 - X_3)$		$(X_4 - X_3)$	$(X_5 - X_3)$
2.025E-07	4.84E-08		5.29E-08	2.116E-07
7.84E-08	1.96E-08		1.96E-08	7.84E-08
8.1E-09	2.5E-09		1.6E-09	6.4E-09
1.44E-08	3.6E-09		3.6E-09	1.21E-08
9.61E-08	2.25E-08		2.56E-08	9.61E-08
2.304E-07	5.76E-08		5.76E-08	2.401E-07
3.844E-07	9.61E-08		1.024E-07	3.969E-07
4.9E-07	1.225E-07		1.225E-07	4.9E-07
5.184E-07	1.296E-07		1.296E-07	5.041E-07
4.489E-07	1.156E-07		1.089E-07	4.356E-07
3.136E-07	7.84E-08		7.84E-08	3.249E-07
1.936E-07	4.84E-08		4.84E-08	1.936E-07
9E-08	2.25E-08		2.25E-08	9E-08
3.24E-08	8.1E-09		8.1E-09	3.24E-08
8.1E-09	2.5E-09		1.6E-09	8.1E-09
4E-10	1E-10		1E-10	4E-10
9E-10	4E-10		1E-10	4E-10
1.6E-09	4E-10		4E-10	2.5E-09
1.6E-09	4E-10		4E-10	1.6E-09
1E-10	1E-10		1E-10	1E-10
4E-10	1E-10		1E-10	9E-10
3.6E-09	9E-10		4E-10	2.5E-09
3.6E-09	9E-10		9E-10	3.6E-09
1.6E-09	4E-10		9E-10	2.5E-09
1E-10	1E-10		1E-10	4E-10
4E-10	1E-10		1E-10	1E-10
1.6E-09	4E-10		4E-10	1.6E-09
2.5E-09	4E-10		4E-10	1.6E-09
1E-10	1E-10		0	0
4.9E-09	1.6E-09		9E-10	3.6E-09
1.96E-08	4.9E-09		4.9E-09	1.96E-08
4.41E-08	1.21E-08		1.21E-08	4.84E-08
7.84E-08	1.96E-08		1.96E-08	7.84E-08
1.024E-07	2.56E-08		2.56E-08	1.024E-07
9.61E-08	2.25E-08		2.56E-08	1.024E-07
7.84E-08	1.96E-08		1.96E-08	7.84E-08
4.41E-08	1E-08		1E-08	4E-08
8.1E-09	1.6E-09		2.5E-09	1E-08
4E-10	1E-10		1E-10	1E-10
1.69E-08	4.9E-09		3.6E-09	1.44E-08
4E-08	1E-08		1.21E-08	4.41E-08
7.29E-08	1.96E-08		1.69E-08	6.76E-08
7.84E-08	1.96E-08		1.96E-08	7.84E-08

6.76E-08	1.69E-08		1.96E-08	7.29E-08
5.29E-08	1.21E-08		1.21E-08	4.84E-08
2.89E-08	8.1E-09		6.4E-09	2.56E-08
1E-08	2.5E-09		2.5E-09	1.21E-08
3.6E-09	9E-10		4E-10	2.5E-09
9E-10	4E-10		1E-10	4E-10
1E-10	0		1E-10	4E-10
9E-10	4E-10		1E-10	9E-10
6.4E-09	1.6E-09		1.6E-09	6.4E-09
2.25E-08	4.9E-09		6.4E-09	2.56E-08
6.25E-08	1.69E-08		1.44E-08	5.76E-08
1.296E-07	3.24E-08		3.24E-08	1.296E-07
2.304E-07	5.76E-08		5.76E-08	2.401E-07
3.844E-07	9.61E-08		9.61E-08	3.969E-07
5.625E-07	1.444E-07		1.369E-07	5.625E-07
7.225E-07	1.849E-07		1.764E-07	7.225E-07
8.281E-07	2.116E-07		2.025E-07	8.1E-07
8.649E-07	2.116E-07		2.209E-07	8.649E-07
8.649E-07	2.209E-07		2.116E-07	8.464E-07
8.836E-07	2.209E-07		2.116E-07	8.649E-07
9.604E-07	2.401E-07		2.401E-07	9.604E-07
1.166E-06	2.916E-07		2.916E-07	1.145E-06
1.513E-06	3.844E-07		3.721E-07	1.488E-06
1.932E-06	4.9E-07		4.9E-07	1.932E-06
2.403E-06	6.084E-07		6.084E-07	2.403E-06
2.756E-06	6.889E-07		6.724E-07	2.723E-06
2.856E-06	7.225E-07		7.056E-07	2.822E-06
2.723E-06	6.889E-07		6.724E-07	2.69E-06
2.434E-06	6.084E-07		5.929E-07	2.403E-06
2.074E-06	5.184E-07		5.184E-07	2.103E-06
1.769E-06	4.356E-07		4.489E-07	1.796E-06
1.488E-06	3.721E-07		3.721E-07	1.488E-06
1.254E-06	3.136E-07		3.136E-07	1.232E-06
1.02E-06	2.5E-07		2.601E-07	1.04E-06
8.1E-07	2.025E-07		2.025E-07	7.921E-07
5.929E-07	1.521E-07		1.521E-07	5.929E-07
3.969E-07	9.61E-08		1.024E-07	3.969E-07
2.304E-07	5.76E-08		5.76E-08	2.304E-07
1.089E-07	2.56E-08		2.89E-08	1.089E-07
3.24E-08	8.1E-09		8.1E-09	3.24E-08
2.5E-09	4E-10		4E-10	2.5E-09
3.6E-09	9E-10		1.6E-09	4.9E-09
1.96E-08	4.9E-09		4.9E-09	1.96E-08
3.24E-08	8.1E-09		8.1E-09	3.24E-08
2.89E-08	6.4E-09		8.1E-09	2.89E-08
1.21E-08	3.6E-09		2.5E-09	1.21E-08
1E-10	1E-10		0	0
2.56E-08	6.4E-09		8.1E-09	2.89E-08
1.296E-07	3.24E-08		3.24E-08	1.296E-07

3.136E-07	7.84E-08		8.41E-08	3.249E-07
5.776E-07	1.444E-07		1.521E-07	5.929E-07
8.836E-07	2.209E-07		2.209E-07	8.649E-07
1.103E-06	2.704E-07		2.704E-07	1.082E-06
1.145E-06	2.916E-07		2.916E-07	1.145E-06
1.04E-06	2.601E-07		2.601E-07	1.04E-06
7.921E-07	1.936E-07		2.025E-07	8.1E-07
5.184E-07	1.296E-07		1.296E-07	5.184E-07
2.601E-07	6.25E-08		6.76E-08	2.704E-07

RMS	RMS		RMS	RMS
0.0067042	0.0033569		0.0033505	0.0066934

Min X	-0.06456
Max X	0.11123
Range X	0.17579

% Error	% Error	% Error	% Error
3.8137551	1.9096354	1.9059546	3.8076325

% Error = (RMS / Range) x 100%

Table E3: Calculation or percent error for moments at the knee joint (LM axis)

Y +0.02kg/m	Y +0.01kg/m	Y Original	Y -0.01kg/m	Y -0.02kg/m
(Y1 – Y3)	(Y2 – Y3)		(Y4 – Y3)	(Y5 – Y3)
1.176E-05	2.958E-06		2.958E-06	1.176E-05
9.181E-06	2.28E-06		2.28E-06	9.12E-06
6.502E-06	1.613E-06		1.638E-06	6.554E-06
4.202E-06	1.04E-06		1.061E-06	4.244E-06
2.434E-06	6.084E-07		6.084E-07	2.465E-06
1.166E-06	2.916E-07		2.916E-07	1.188E-06
3.969E-07	1.024E-07		9.61E-08	3.844E-07
4.41E-08	1E-08		1.21E-08	4.41E-08
1.44E-08	3.6E-09		2.5E-09	1.21E-08
9.61E-08	2.25E-08		2.56E-08	1.024E-07
1.369E-07	3.61E-08		3.61E-08	1.369E-07
1.024E-07	2.56E-08		2.56E-08	1.024E-07
4.41E-08	1E-08		1.21E-08	4.41E-08
8.1E-09	1.6E-09		2.5E-09	8.1E-09
4E-10	1E-10		1E-10	4E-10
4E-10	1E-10		1E-10	4E-10
8.1E-09	2.5E-09		1.6E-09	8.1E-09
4E-08	1E-08		1E-08	4E-08
9E-08	2.25E-08		2.25E-08	9.61E-08
1.521E-07	3.61E-08		4E-08	1.521E-07
1.849E-07	4.84E-08		4.41E-08	1.849E-07
1.936E-07	4.84E-08		4.84E-08	1.936E-07
2.025E-07	5.29E-08		4.84E-08	1.936E-07
2.209E-07	5.76E-08		5.29E-08	2.209E-07
3.025E-07	7.29E-08		7.29E-08	2.916E-07
4.624E-07	1.156E-07		1.156E-07	4.624E-07
7.744E-07	1.936E-07		1.849E-07	7.569E-07
1.21E-06	3.025E-07		3.025E-07	1.21E-06
1.69E-06	4.225E-07		4.356E-07	1.716E-06
2.102E-06	5.329E-07		5.329E-07	2.103E-06
2.28E-06	5.776E-07		5.625E-07	2.25E-06
2.074E-06	5.184E-07		5.041E-07	2.045E-06
1.563E-06	3.844E-07		3.969E-07	1.563E-06
9.216E-07	2.304E-07		2.401E-07	9.409E-07
3.844E-07	9.61E-08		9.61E-08	3.844E-07
5.29E-08	1.21E-08		1.44E-08	5.76E-08
1.96E-08	4.9E-09		6.4E-09	2.25E-08
2.601E-07	6.25E-08		6.76E-08	2.601E-07
6.889E-07	1.764E-07		1.764E-07	7.056E-07
1.254E-06	3.136E-07		3.025E-07	1.232E-06
1.822E-06	4.624E-07		4.489E-07	1.796E-06
2.28E-06	5.625E-07		5.625E-07	2.25E-06
2.528E-06	6.4E-07		6.4E-07	2.528E-06

2.624E-06	6.561E-07
2.56E-06	6.4E-07
2.434E-06	6.084E-07
2.31E-06	5.776E-07
2.28E-06	5.625E-07
2.341E-06	5.776E-07
2.528E-06	6.4E-07
2.856E-06	7.225E-07
3.24E-06	8.1E-07
3.686E-06	9.216E-07
4.121E-06	1.02E-06
4.368E-06	1.102E-06
4.41E-06	1.102E-06
3.96E-06	0.000001
2.89E-06	7.225E-07
1.464E-06	3.6E-07
2.304E-07	5.76E-08
2.5E-07	6.25E-08
2.993E-06	7.569E-07
9.548E-06	2.372E-06
2.034E-05	5.108E-06
3.411E-05	8.526E-06
4.858E-05	1.218E-05
6.1E-05	1.529E-05
6.939E-05	1.739E-05
7.344E-05	1.84E-05
7.362E-05	1.84E-05
7.123E-05	1.781E-05
6.708E-05	1.673E-05
6.225E-05	1.56E-05
5.655E-05	1.414E-05
5.069E-05	1.267E-05
4.476E-05	1.116E-05
3.869E-05	9.672E-06
3.306E-05	8.237E-06
2.756E-05	6.917E-06
2.247E-05	5.617E-06
1.781E-05	4.452E-06
1.332E-05	3.349E-06
9.303E-06	2.341E-06
5.664E-06	1.416E-06
2.657E-06	6.561E-07
6.084E-07	1.521E-07
2.25E-08	6.4E-09
1.346E-06	3.364E-07
5.153E-06	1.3E-06
1.156E-05	2.89E-06
2.061E-05	5.153E-06
3.192E-05	7.952E-06

6.561E-07	2.592E-06
6.4E-07	2.56E-06
6.084E-07	2.434E-06
5.776E-07	2.31E-06
5.776E-07	2.28E-06
5.929E-07	2.372E-06
6.4E-07	2.56E-06
7.056E-07	2.822E-06
8.1E-07	3.24E-06
9.216E-07	3.686E-06
1.04E-06	4.121E-06
1.102E-06	4.41E-06
1.102E-06	4.41E-06
9.801E-07	3.92E-06
7.396E-07	2.924E-06
3.721E-07	1.464E-06
5.76E-08	2.304E-07
6.25E-08	2.5E-07
7.396E-07	2.993E-06
2.403E-06	9.61E-06
5.063E-06	2.034E-05
8.526E-06	3.411E-05
1.211E-05	4.858E-05
1.521E-05	6.1E-05
1.739E-05	6.956E-05
1.832E-05	7.344E-05
1.84E-05	7.362E-05
1.781E-05	7.123E-05
1.681E-05	6.724E-05
1.552E-05	6.209E-05
1.421E-05	5.67E-05
1.267E-05	5.084E-05
1.116E-05	4.476E-05
9.672E-06	3.881E-05
8.237E-06	3.295E-05
6.917E-06	2.756E-05
5.617E-06	2.256E-05
4.452E-06	1.781E-05
3.349E-06	1.34E-05
2.31E-06	9.303E-06
1.416E-06	5.664E-06
6.561E-07	2.657E-06
1.6E-07	6.241E-07
4.9E-09	1.96E-08
3.481E-07	1.369E-06
1.277E-06	5.108E-06
2.924E-06	1.163E-05
5.198E-06	2.07E-05
8.009E-06	3.204E-05

4.436E-05	1.109E-05		1.109E-05	4.436E-05
5.64E-05	1.406E-05		1.414E-05	5.64E-05
6.626E-05	1.656E-05		1.656E-05	6.626E-05
7.242E-05	1.815E-05		1.806E-05	7.242E-05
7.344E-05	1.84E-05		1.832E-05	7.344E-05
6.872E-05	1.714E-05		1.714E-05	6.872E-05
5.929E-05	1.482E-05		1.475E-05	5.914E-05
4.706E-05	1.176E-05		1.17E-05	4.692E-05
3.399E-05	8.468E-06		8.526E-06	3.411E-05

RMS	RMS		RMS	RMS
0.0402205	0.0201135		0.0201083	0.0402305

Min Y	-0.13428
Max Y	0.29684
Range Y	0.43112

% Error	% Error	% Error	% Error
9.3293079	4.665395	4.6641977	9.3316245

% Error = (RMS / Range) x 100%

Table E4: Calculation or percent error for moments at the knee joint (DP axis)

Z +0.02kg/m	Z +0.01kg/m	Z Original	Z -0.01kg/m	Z -0.02kg/m
(Z1 – Z3)	(Z2 – Z3)		(Z4 – Z3)	(Z5 – Z3)
4E-10	1E-10		0	1E-10
2.5E-09	9E-10		4E-10	2.5E-09
1.69E-08	4.9E-09		3.6E-09	1.44E-08
3.61E-08	1E-08		1E-08	4E-08
6.25E-08	1.44E-08		1.69E-08	6.76E-08
7.84E-08	1.96E-08		1.96E-08	7.84E-08
6.76E-08	1.69E-08		1.96E-08	7.29E-08
5.29E-08	1.44E-08		1.21E-08	4.84E-08
2.25E-08	4.9E-09		4.9E-09	1.96E-08
2.5E-09	9E-10		4E-10	1.6E-09
3.6E-09	9E-10		9E-10	3.6E-09
2.56E-08	6.4E-09		4.9E-09	2.25E-08
4.84E-08	1.21E-08		1.21E-08	4.84E-08
6.76E-08	1.69E-08		1.69E-08	6.76E-08
6.76E-08	1.69E-08		1.44E-08	6.25E-08
5.29E-08	1.21E-08		1.44E-08	5.29E-08
4E-08	1E-08		1E-08	3.61E-08
2.56E-08	6.4E-09		6.4E-09	2.56E-08
1.69E-08	4.9E-09		3.6E-09	1.69E-08
1.21E-08	2.5E-09		2.5E-09	1E-08
8.1E-09	1.6E-09		1.6E-09	8.1E-09
4.9E-09	1.6E-09		1.6E-09	4.9E-09
3.6E-09	9E-10		1.6E-09	4.9E-09
3.6E-09	9E-10		4E-10	2.5E-09
1.6E-09	4E-10		9E-10	2.5E-09
1.6E-09	4E-10		1E-10	9E-10
9E-10	4E-10		1E-10	9E-10
4E-10	1E-10		1E-10	4E-10
0	0		1E-10	1E-10
1E-10	0		0	0
1E-10	1E-10		0	1E-10
1E-10	0		1E-10	4E-10
4E-10	1E-10		1E-10	4E-10
9E-10	4E-10		1E-10	4E-10
9E-10	1E-10		4E-10	9E-10
1.6E-09	4E-10		4E-10	1.6E-09
2.5E-09	9E-10		4E-10	1.6E-09
2.5E-09	4E-10		9E-10	3.6E-09
3.6E-09	9E-10		9E-10	3.6E-09
3.6E-09	9E-10		4E-10	2.5E-09
1.6E-09	4E-10		1E-10	9E-10
0	0		1E-10	1E-10

9E-10	1E-10
2.5E-09	4E-10
4.9E-09	9E-10
4.9E-09	9E-10
4.9E-09	1.6E-09
4.9E-09	9E-10
2.5E-09	4E-10
1E-10	1E-10
9E-10	4E-10
1E-08	2.5E-09
3.24E-08	8.1E-09
7.84E-08	1.96E-08
1.444E-07	3.61E-08
2.5E-07	6.25E-08
3.364E-07	8.41E-08
4.096E-07	1.024E-07
4.225E-07	1.089E-07
3.6E-07	9E-08
2.401E-07	6.25E-08
1.024E-07	2.56E-08
1E-08	2.5E-09
2.25E-08	4.9E-09
1.6E-07	4E-08
3.844E-07	9.61E-08
6.241E-07	1.6E-07
7.569E-07	1.849E-07
7.744E-07	1.936E-07
6.561E-07	1.6E-07
4.761E-07	1.156E-07
3.136E-07	7.84E-08
1.764E-07	4.41E-08
8.41E-08	1.96E-08
4.41E-08	1E-08
2.25E-08	6.4E-09
1.21E-08	3.6E-09
6.4E-09	1.6E-09
3.6E-09	9E-10
1.6E-09	4E-10
4E-10	1E-10
0	0
1E-10	0
4E-10	1E-10
9E-10	4E-10
9E-10	1E-10
2.5E-09	4E-10
4.9E-09	1.6E-09
8.1E-09	2.5E-09
1.69E-08	4.9E-09
2.89E-08	8.1E-09

1E-10	4E-10
9E-10	2.5E-09
9E-10	3.6E-09
1.6E-09	4.9E-09
9E-10	4.9E-09
9E-10	3.6E-09
4E-10	2.5E-09
1E-10	4E-10
1E-10	9E-10
1.6E-09	8.1E-09
8.1E-09	3.61E-08
1.96E-08	7.84E-08
4E-08	1.521E-07
5.76E-08	2.401E-07
8.41E-08	3.364E-07
1.024E-07	3.969E-07
1.024E-07	4.096E-07
9.61E-08	3.721E-07
6.25E-08	2.5E-07
2.56E-08	1.024E-07
2.5E-09	1E-08
6.4E-09	2.25E-08
4E-08	1.681E-07
9.61E-08	3.844E-07
1.521E-07	6.084E-07
1.936E-07	7.744E-07
1.936E-07	7.744E-07
1.681E-07	6.724E-07
1.225E-07	4.9E-07
7.29E-08	3.025E-07
4E-08	1.681E-07
2.25E-08	9E-08
1E-08	4.41E-08
4.9E-09	1.96E-08
2.5E-09	1E-08
1.6E-09	6.4E-09
9E-10	3.6E-09
4E-10	1.6E-09
1E-10	4E-10
0	1E-10
0	1E-10
1E-10	4E-10
1E-10	9E-10
4E-10	1.6E-09
9E-10	2.5E-09
9E-10	3.6E-09
2.5E-09	1E-08
3.6E-09	1.69E-08
8.1E-09	2.89E-08

4.84E-08	1.21E-08		1.21E-08	4.84E-08
6.76E-08	1.69E-08		1.69E-08	6.25E-08
8.41E-08	1.96E-08		1.96E-08	8.41E-08
8.41E-08	1.96E-08		2.25E-08	9E-08
8.41E-08	1.96E-08		2.25E-08	9E-08
7.29E-08	1.69E-08		1.96E-08	7.29E-08
5.29E-08	1.44E-08		1.21E-08	4.84E-08
2.89E-08	6.4E-09		8.1E-09	2.89E-08
1.21E-08	3.6E-09		2.5E-09	1E-08
1.6E-09	4E-10		4E-10	1.6E-09

RMS	RMS		RMS	RMS
0.0028788	0.0014371		0.00144	0.0028795

Min Z	-0.01799
Max Z	0.03758
Range Z	0.05557

% Error	% Error	% Error	% Error
5.1804964	2.5861349	2.5912638	5.1817464

% Error = (RMS / Range) x 100%

Appendix F

Time Distance Parameter Graphs

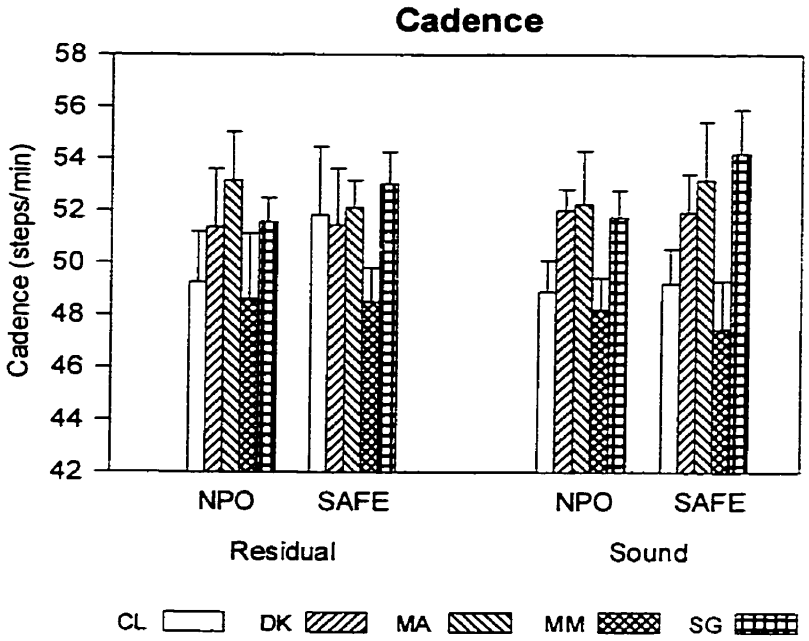


Figure F1: Cadence for five subjects, average of five trials per subject

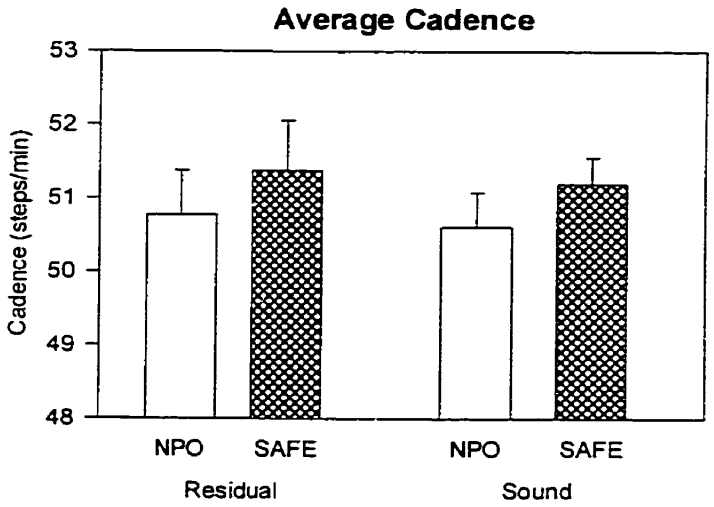


Figure F2: Cadence for five subjects, averaged by prosthetic

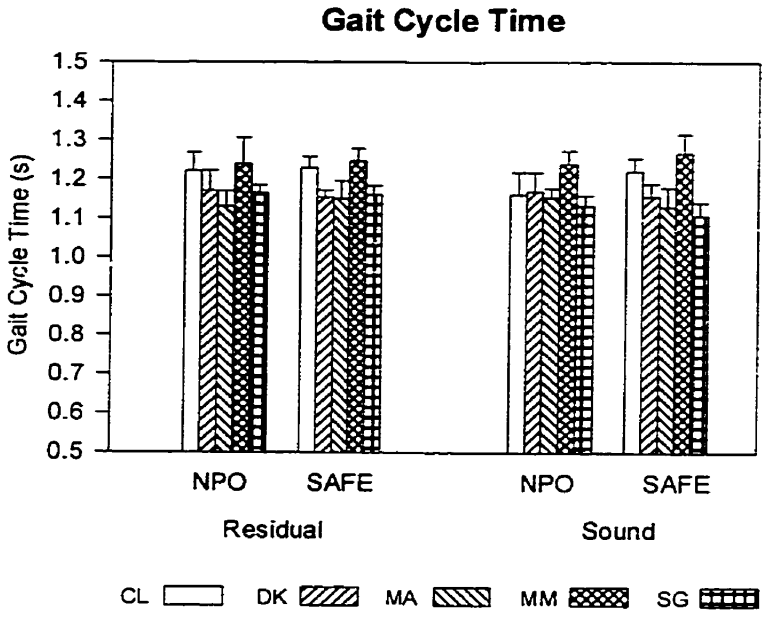


Figure F3: Gait cycle time for five subjects, five trials per subject

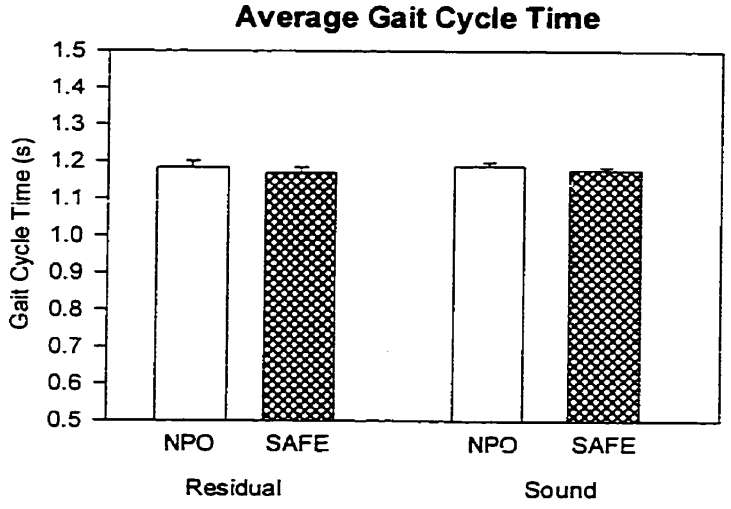


Figure F4: Gait cycle time for five subjects, averaged by prosthetic

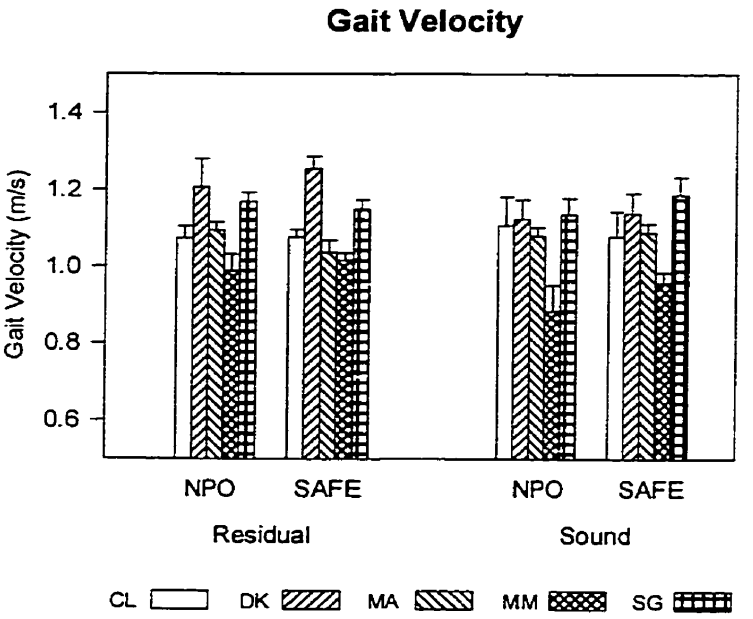


Figure F5: Gait velocity for five subjects, five trials per subject

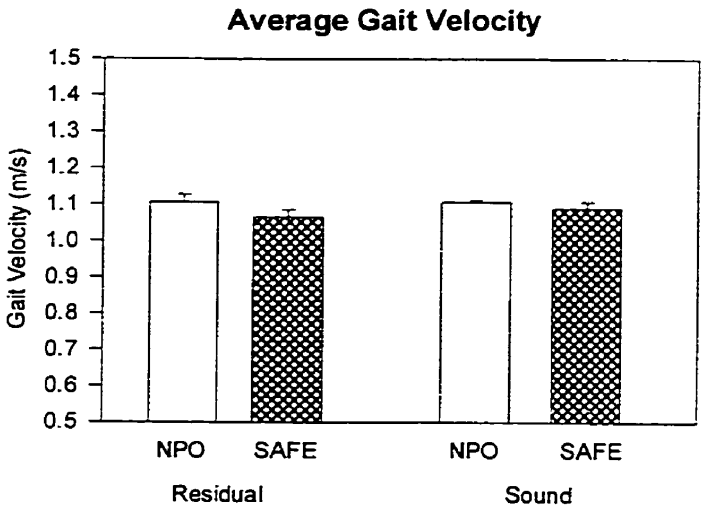


Figure F6: Gait velocity for five subjects, averaged by prosthetic

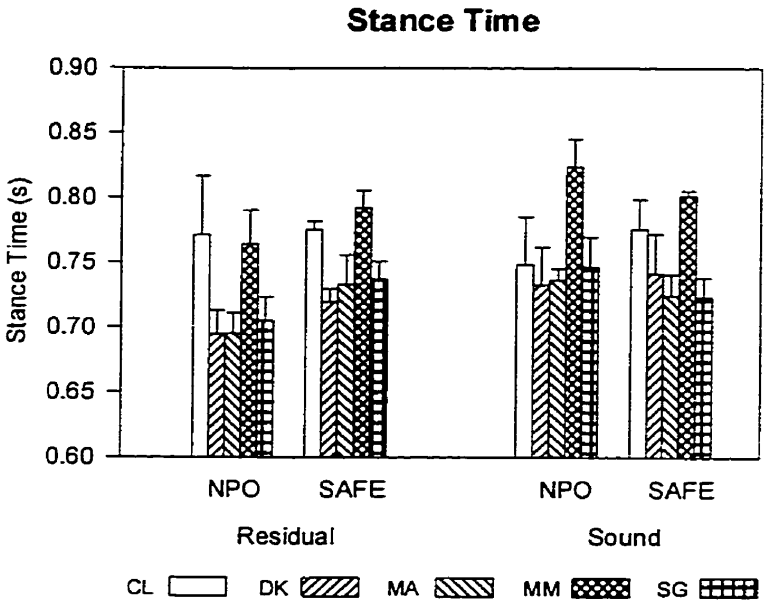


Figure F7: Stance time for five subjects, five trials per subject

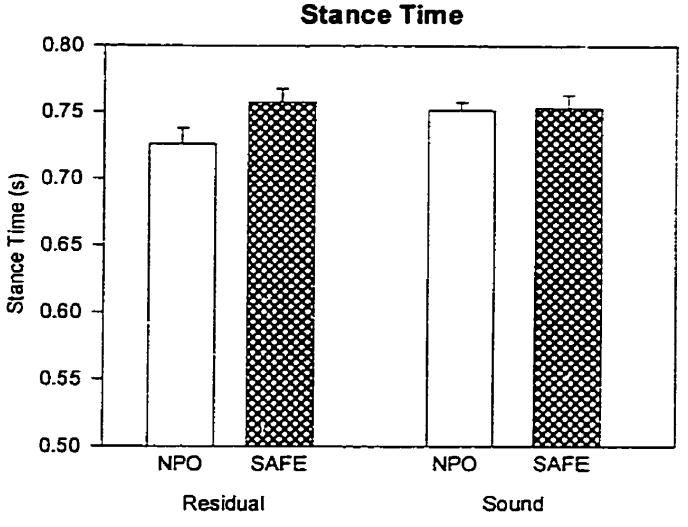


Figure F8: Stance time for five subjects, averaged by prosthetic

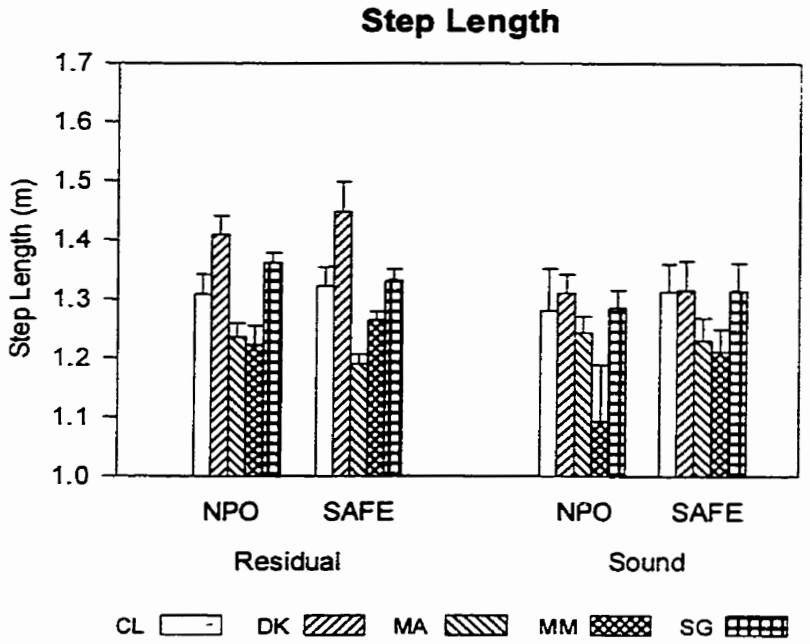


Figure F9: Step length for five subjects, five trials per subject

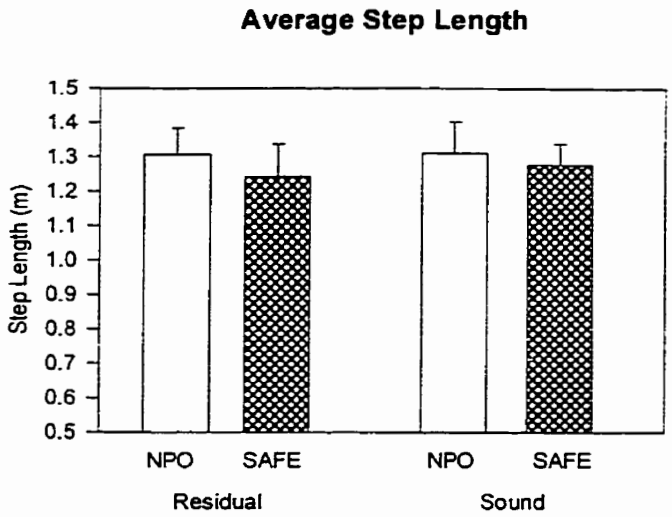


Figure F10: Step length for five subjects, averaged by prosthetic

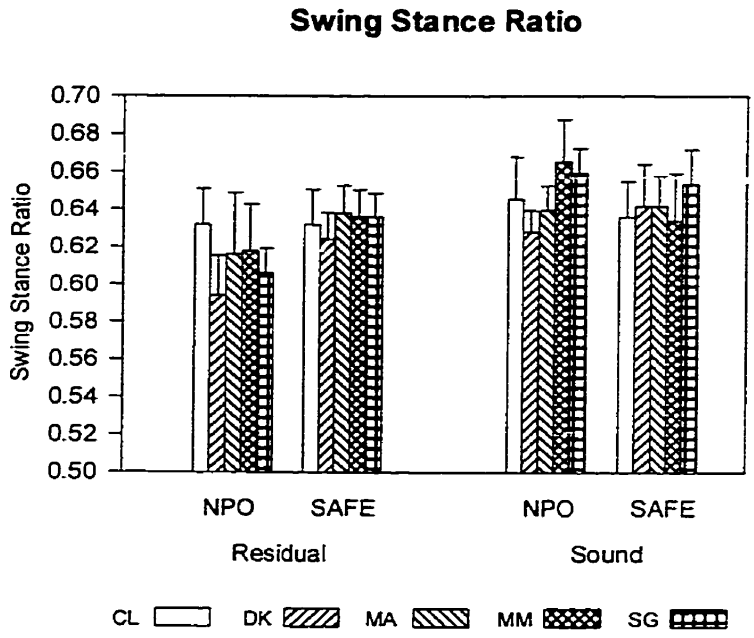


Figure F11: Stance ratio for five subjects, five trials per subject

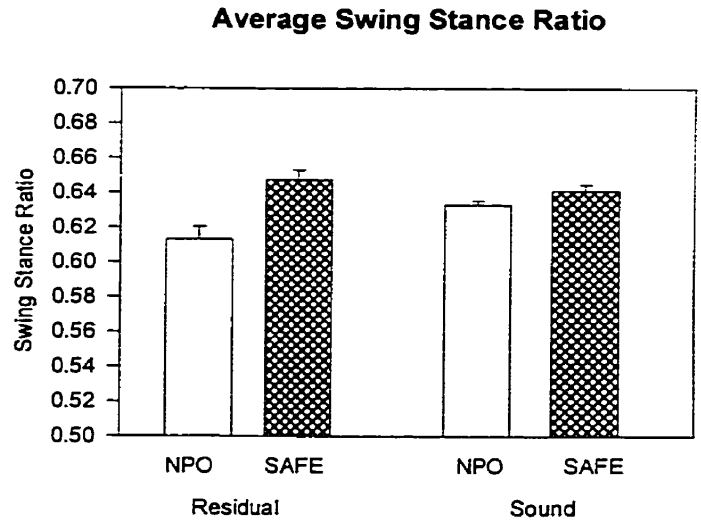
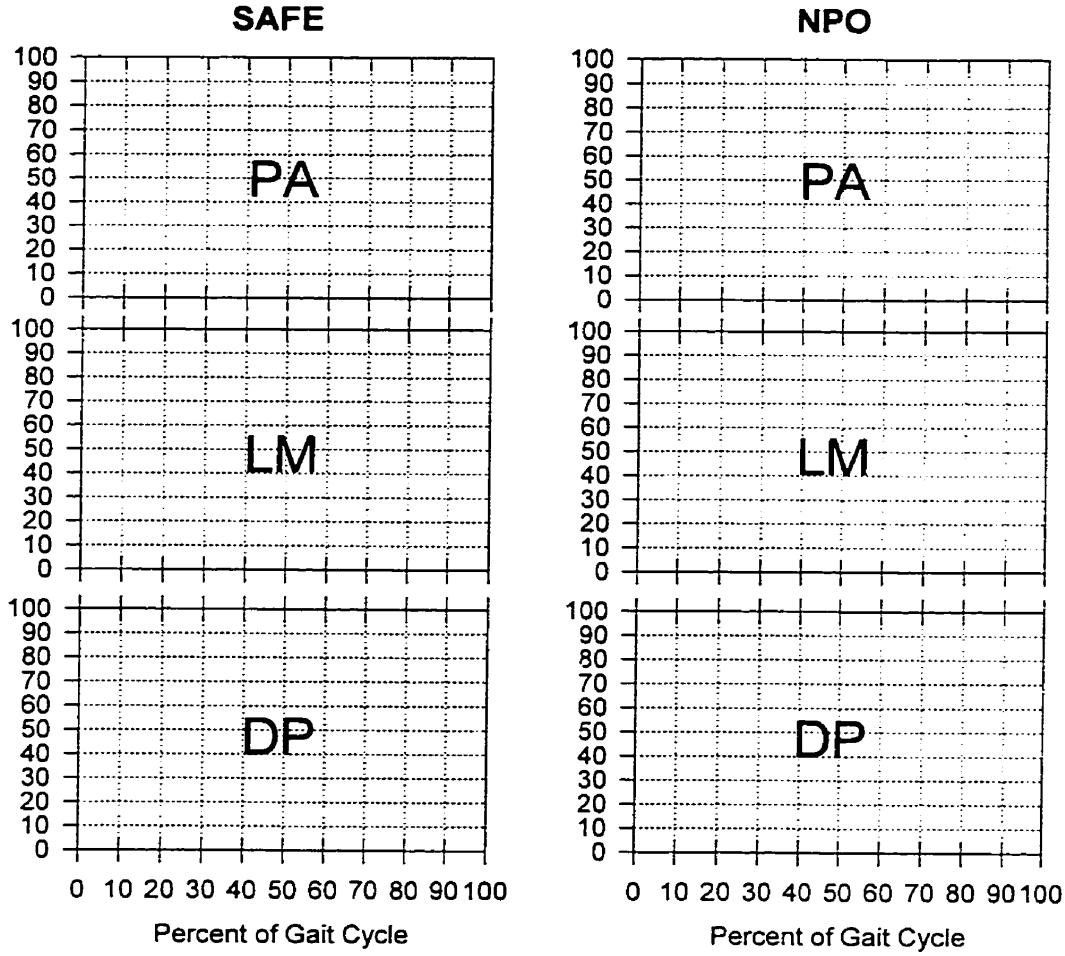


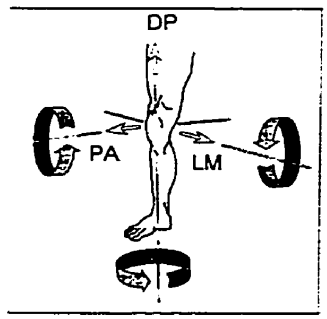
Figure F12: Stance ratio for five subjects, averaged by prosthetic

Appendix G

Averaged Gait Variable Graphs



PA – Posterior / Anterior Axis
LM – Lateral / Medial Axis
DP – Distal / Proximal Axis



By convention Forces are reported along the axis, moments and angles are provided about the axis and conform to the right hand rule.

All graphs in the proceeding appendix follow the format reported above. *
* - Ankle and Metatarsal joint data collected in two dimensions. Only PA data provided.

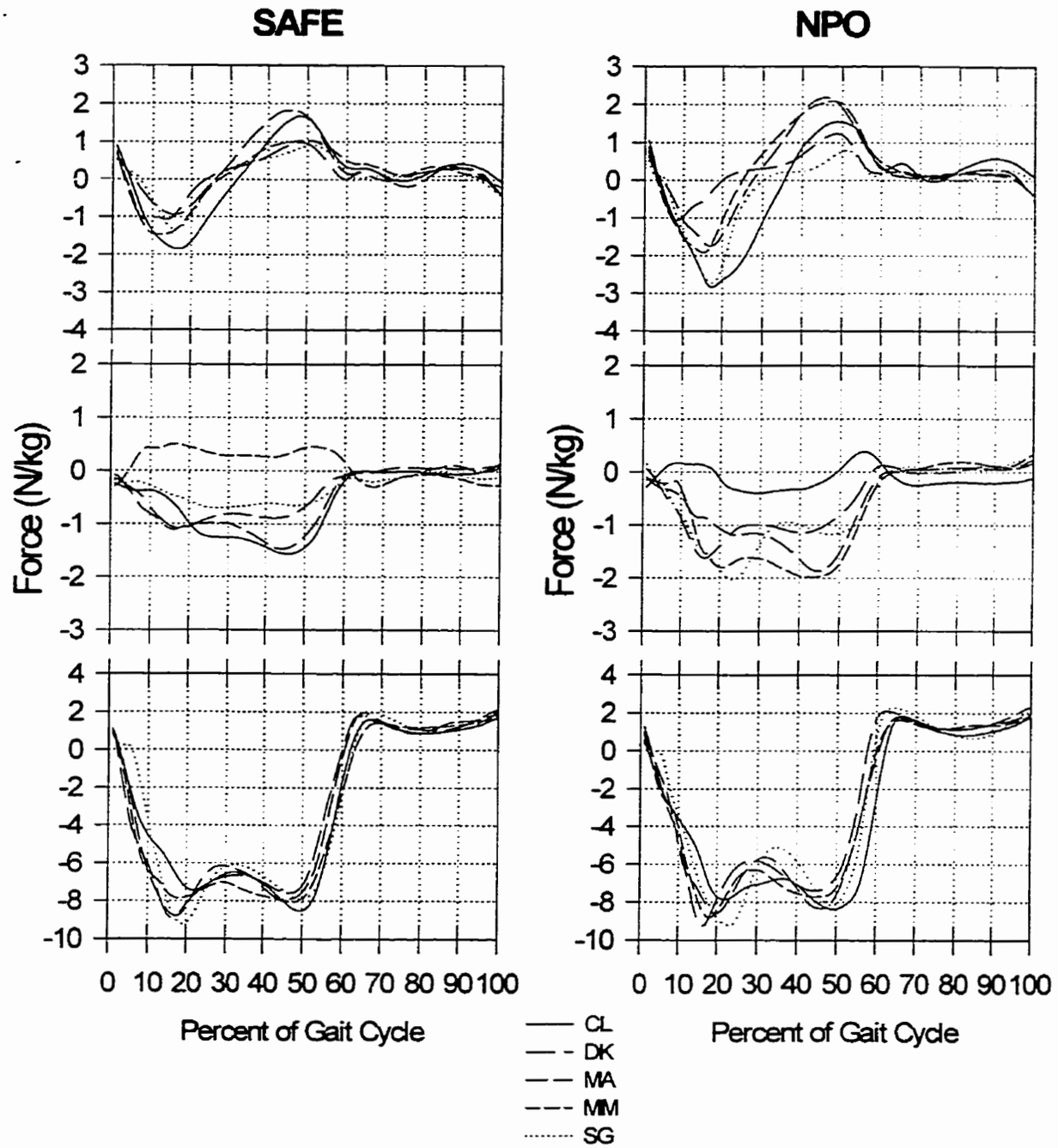


Figure G1: Average Fixed Body Force at the Hip Joint - Residual Limb
Five subjects
Average for five trials per subject
By prosthetic

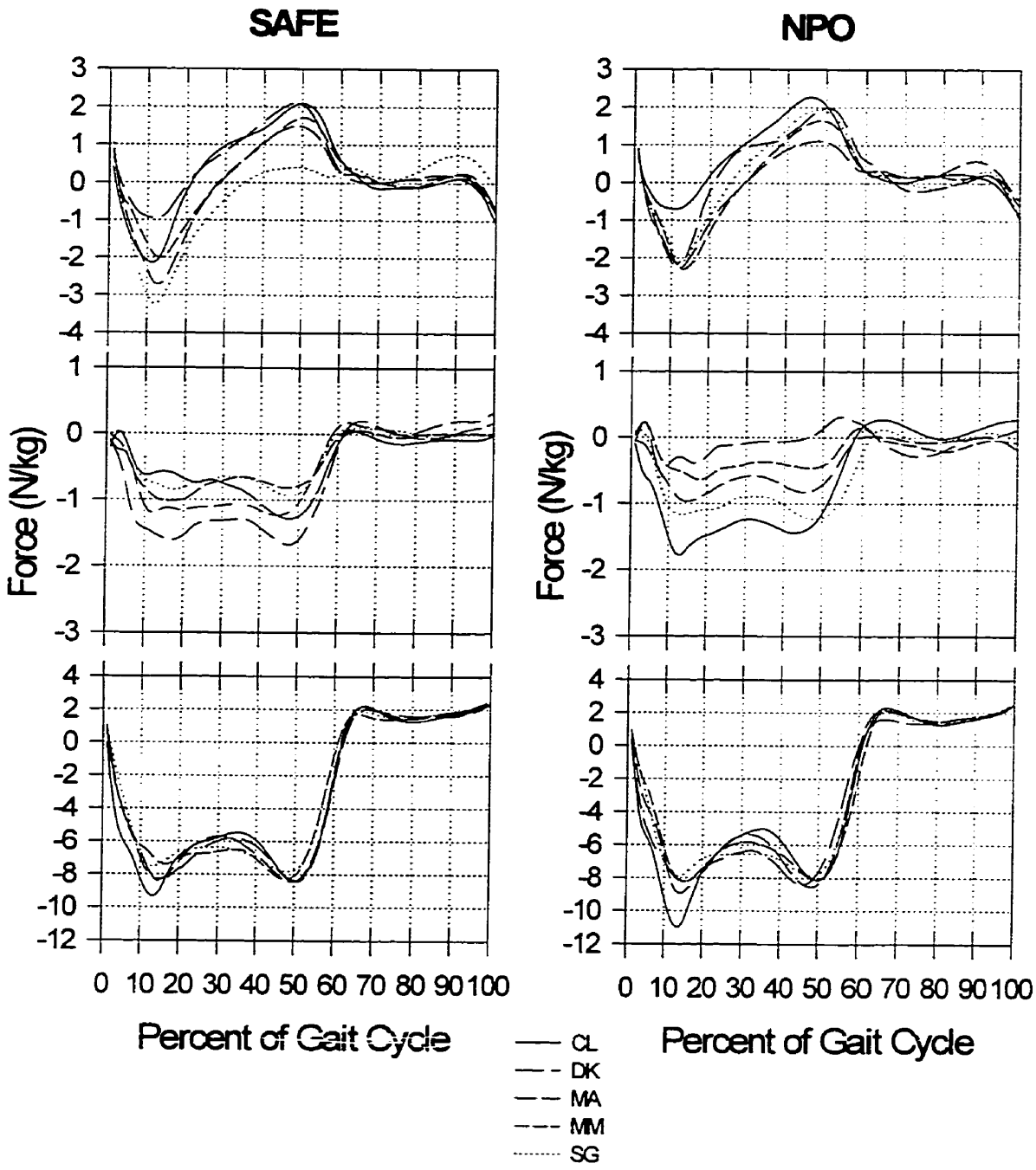


Figure G2: Average Fixed Body Force at the Hip Joint - Sound Limb
Five subjects
Average for five trials per subject
By prosthetic

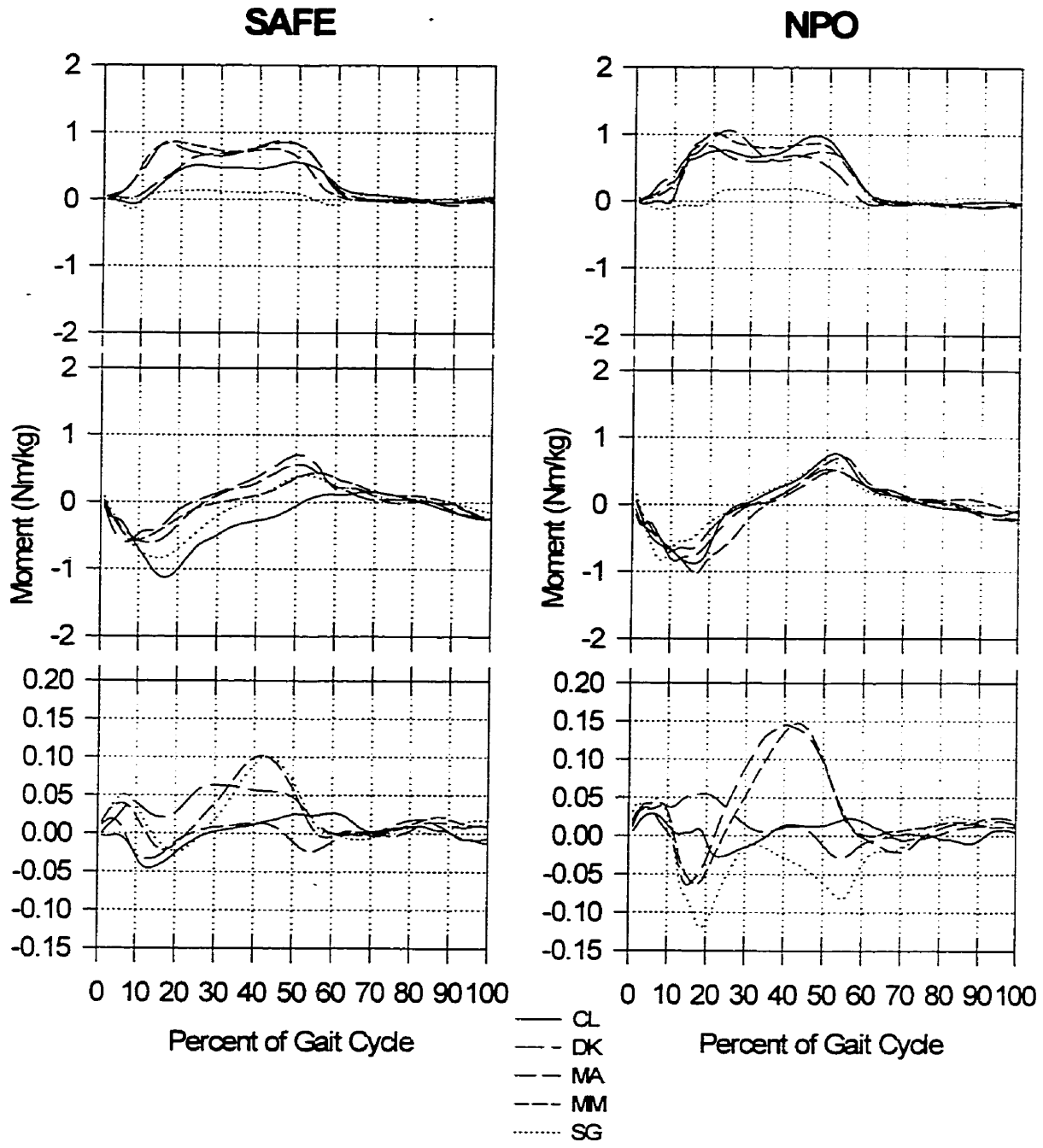


Figure G3: Average Fixed Body Moment at the Hip Joint - Residual Limb
Five subjects
Average for five trials per subject
By prosthetic

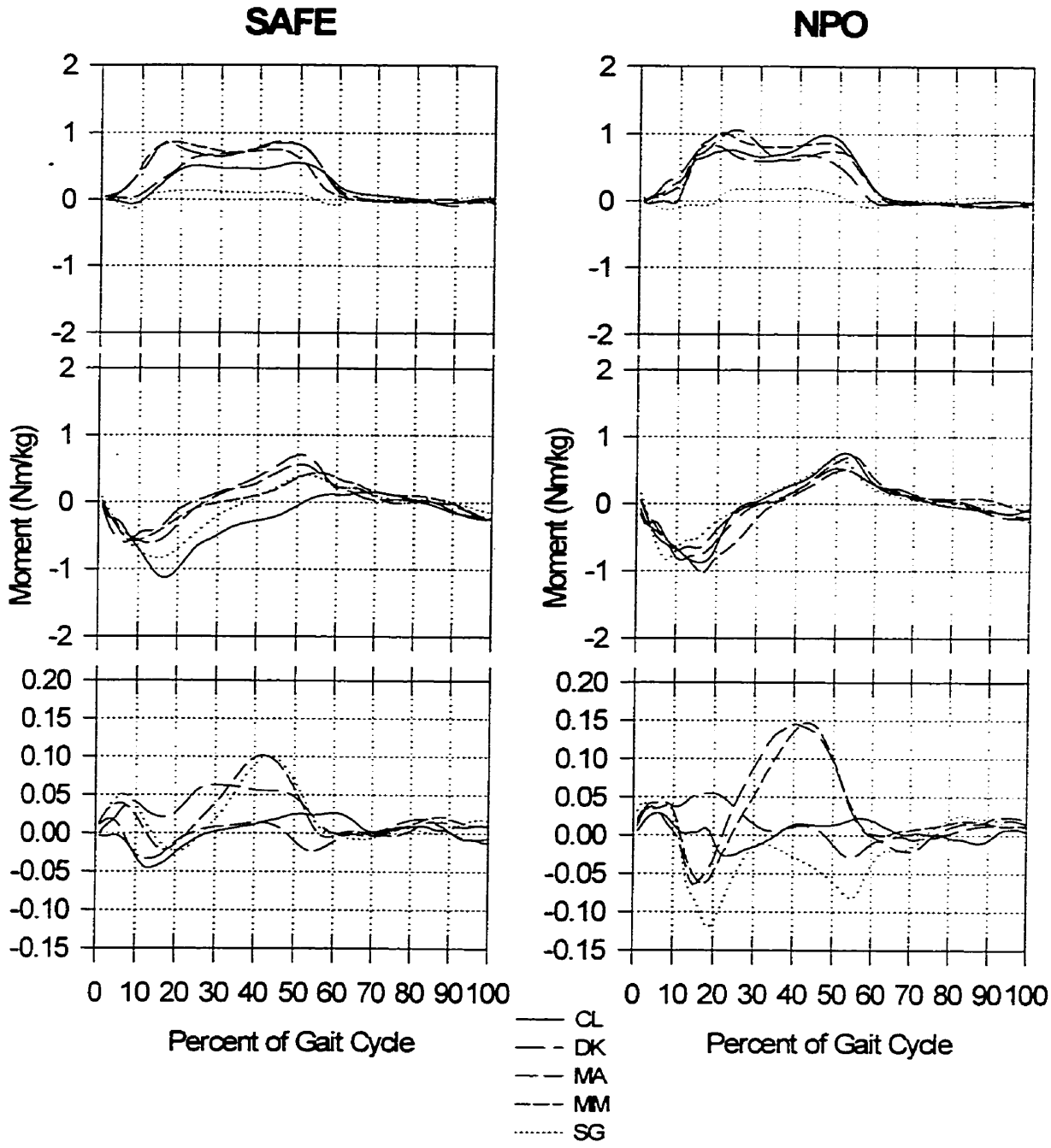


Figure G4: Average Fixed Body Moment at the Hip Joint - Sound Limb
Five subjects
Average for five trials per subject
By prosthetic

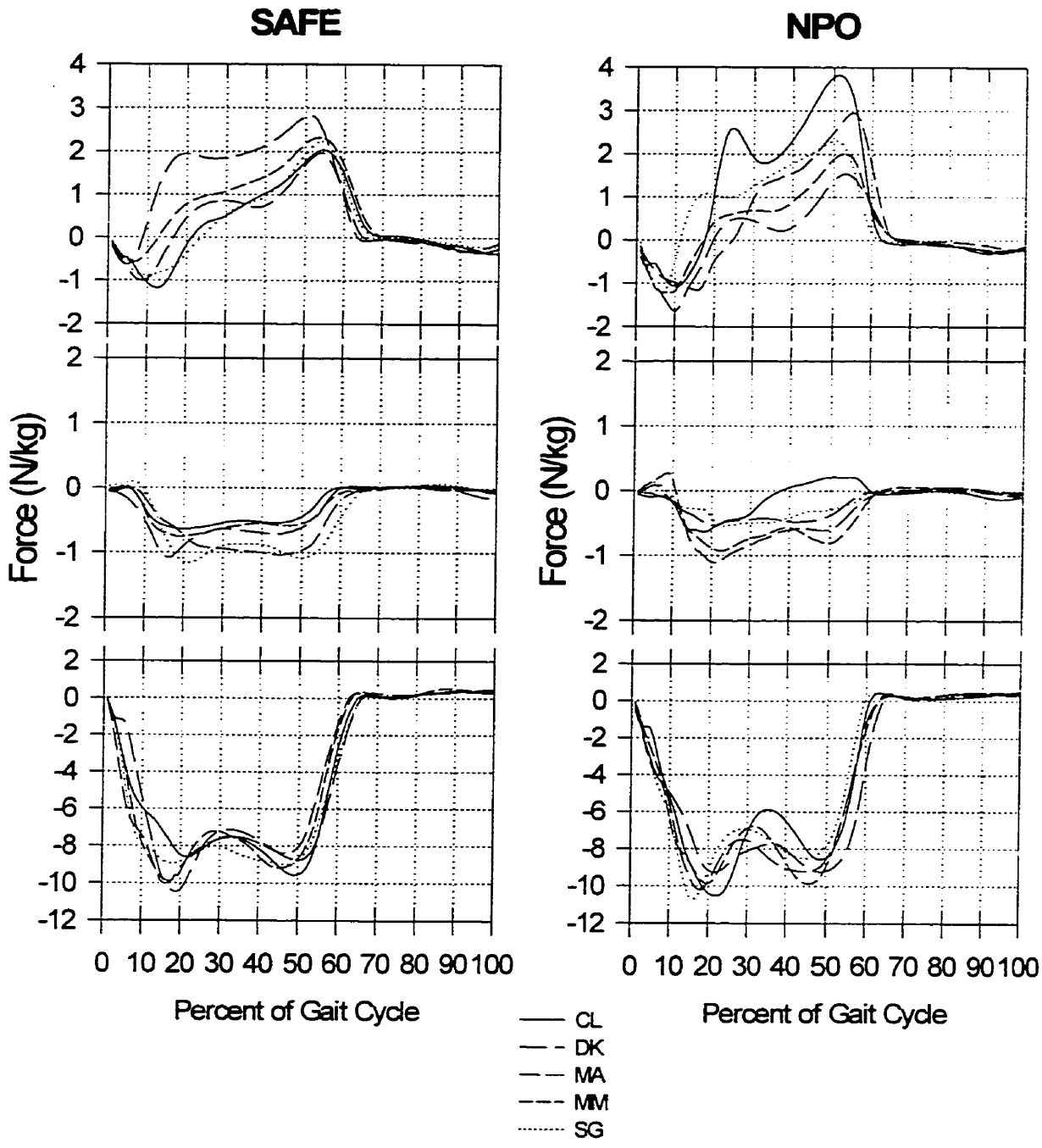


Figure G5: Average Fixed Body Force at the Knee Joint - Residual Limb
Five subjects
Average for five trials per subject
By prosthetic

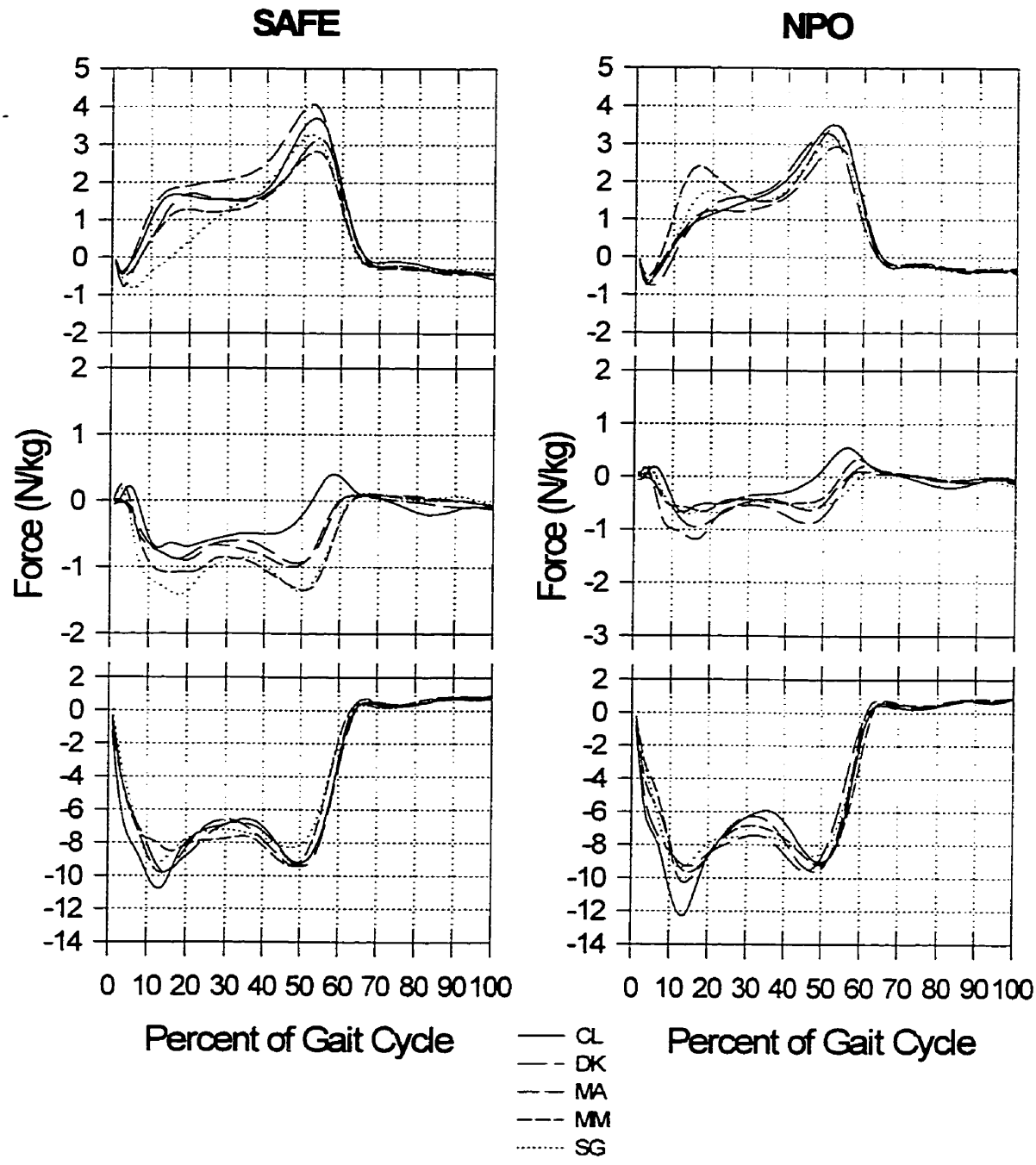


Figure G6: Average Fixed Body Force at the Knee Joint - Sound Limb
Five subjects
Average for five trials per subject
By prosthetic

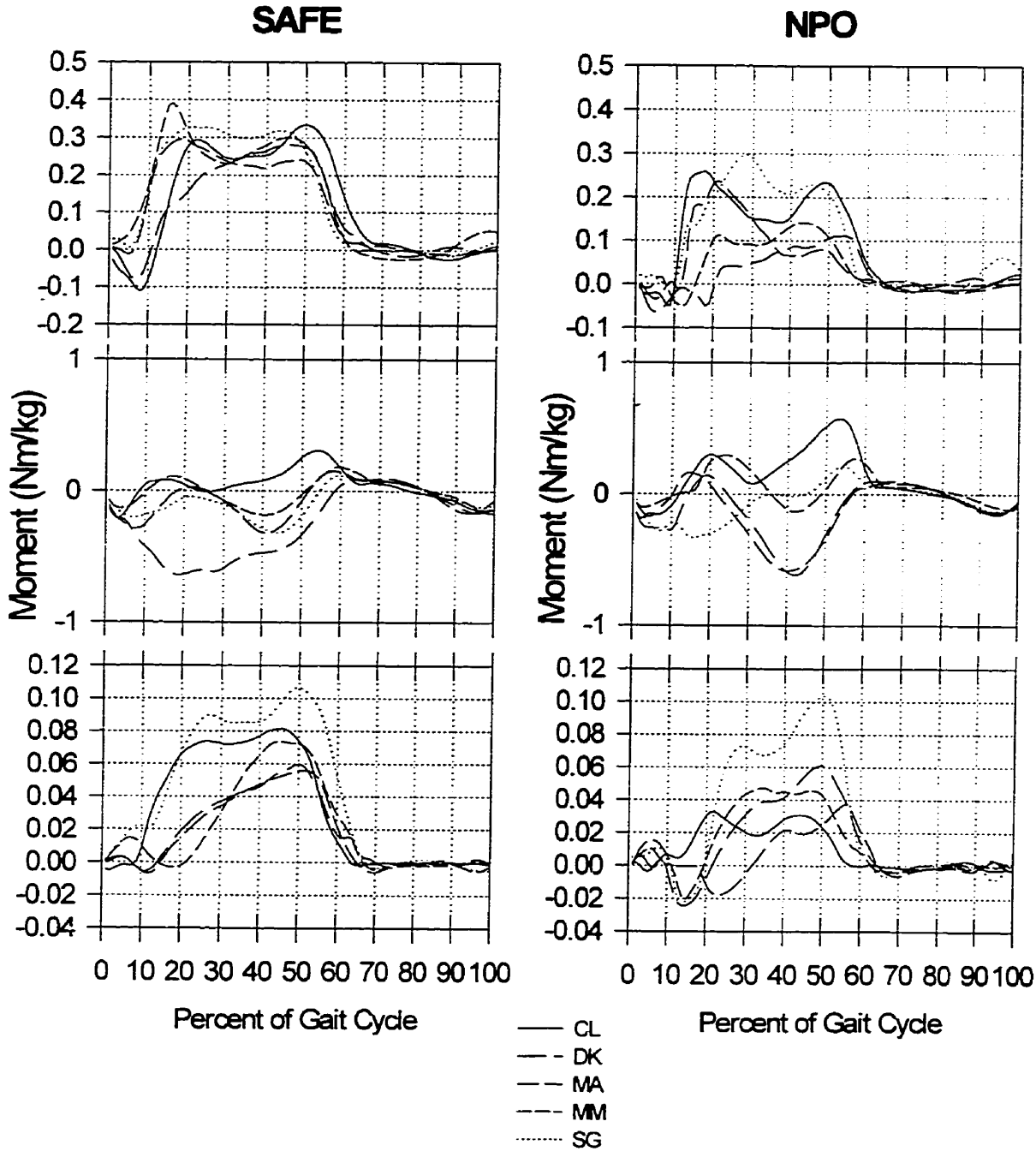


Figure G7: Average Fixed Body Moment at the Knee Joint - Residual Limb
Five subjects
Average for five trials per subject
By prosthetic

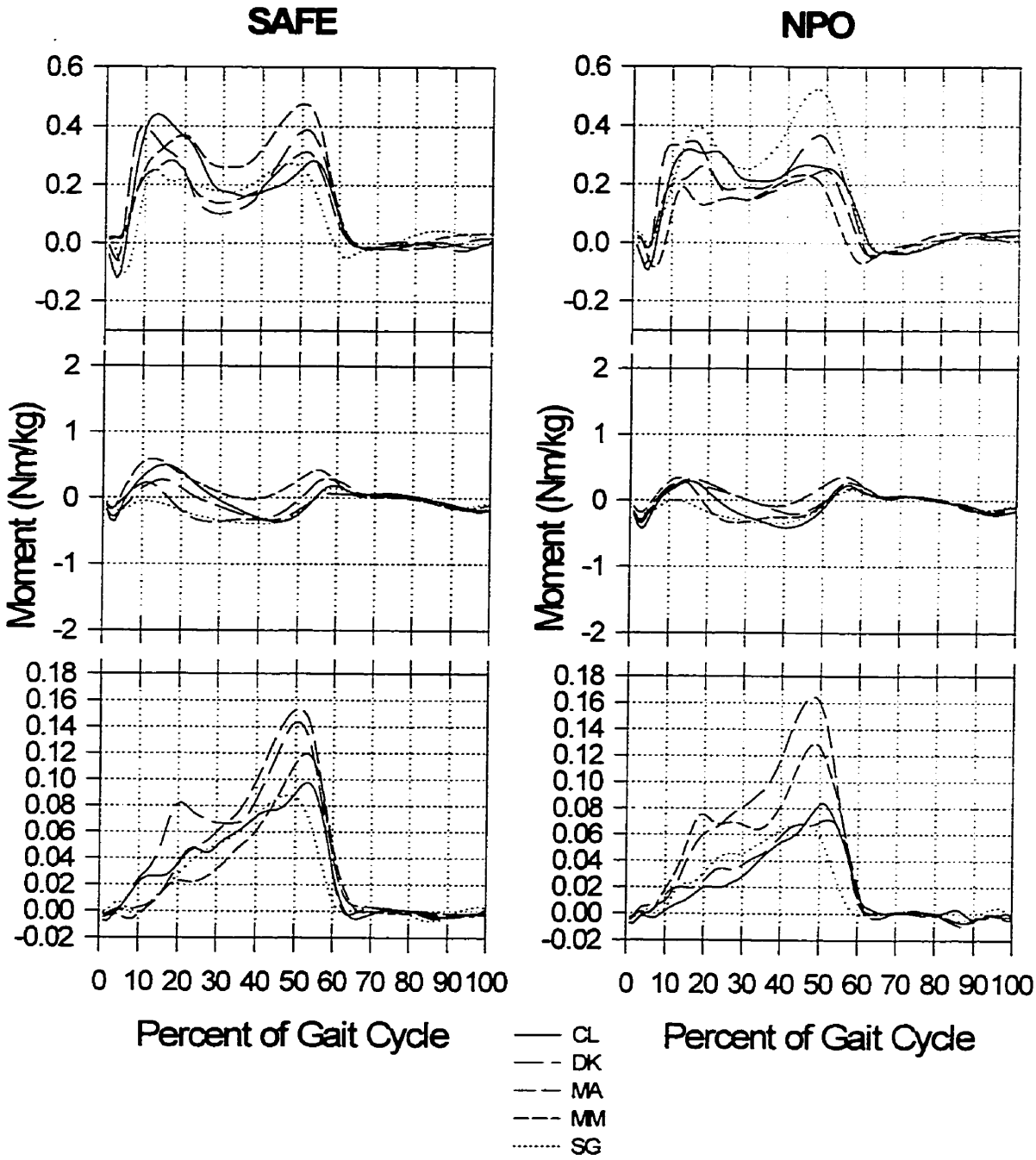


Figure G8: Average Fixed Body Moment at the Knee Joint - Sound Limb
Five subjects
Average for five trials per subject
By prosthetic

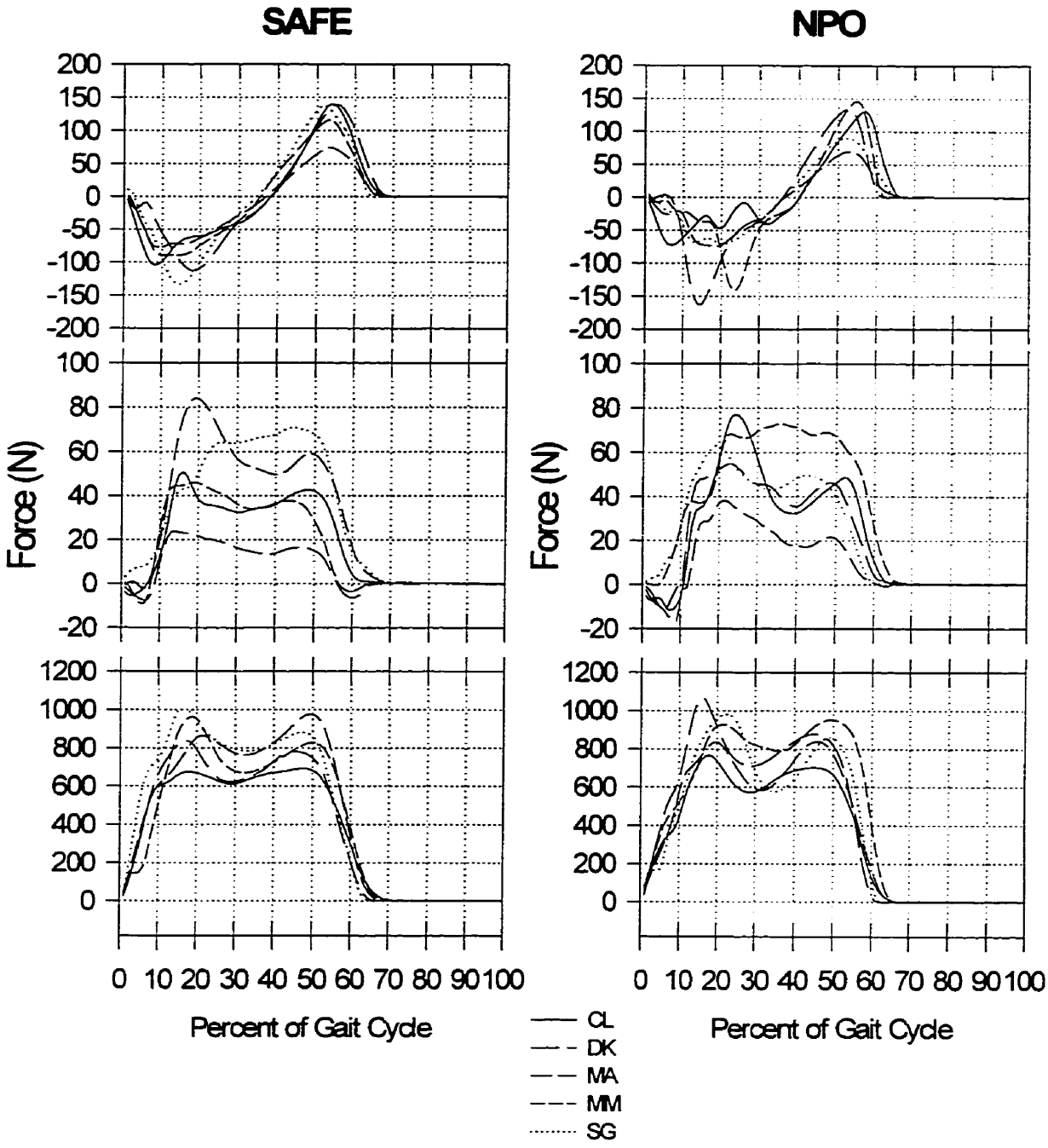


Figure G9: Average Ground Reaction Force - Residual Limb
Five subjects
Average for five trials per subject
By prosthetic

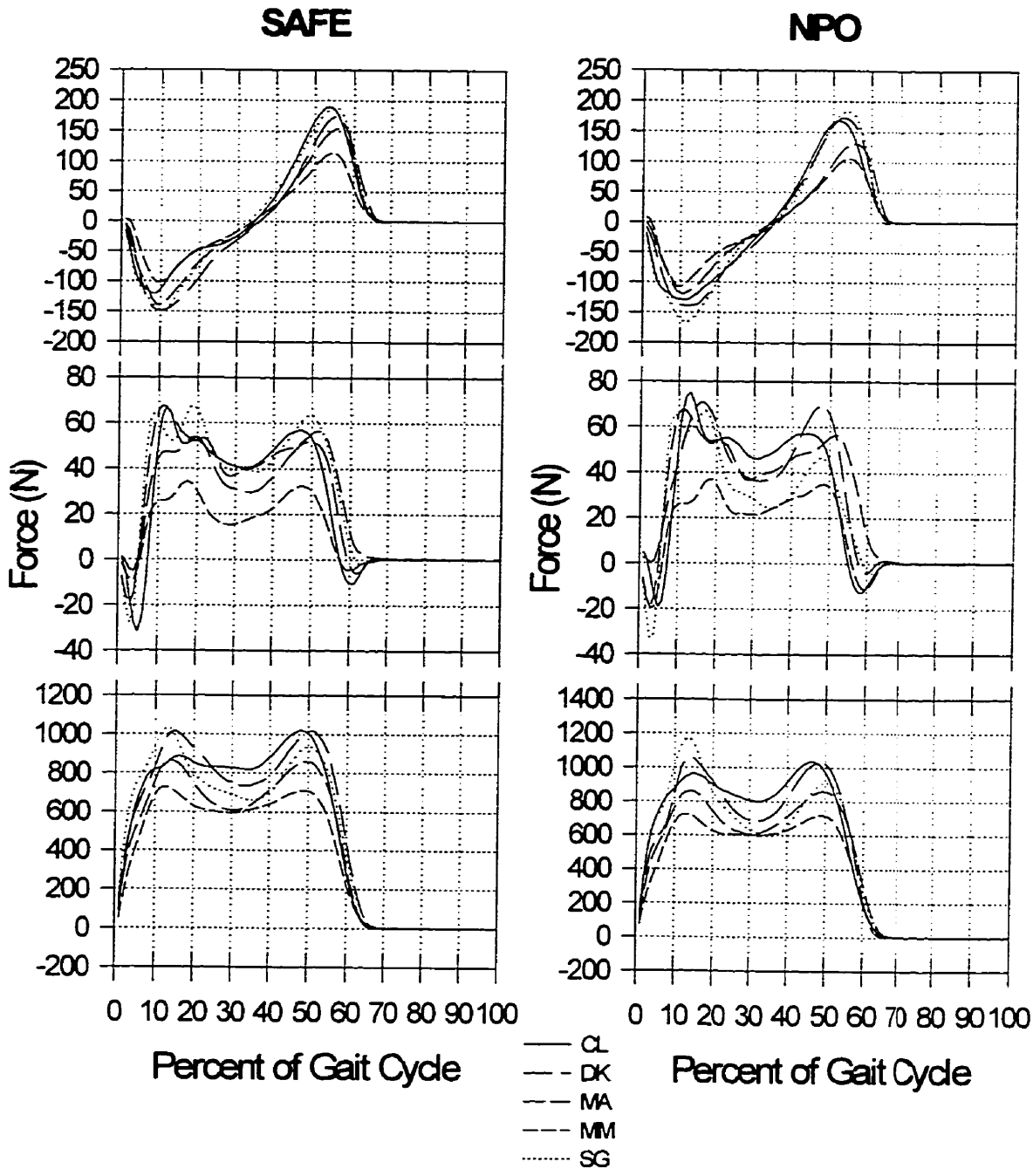


Figure G10: Average Ground Reaction Force - Sound Limb
Five subjects
Average for five trials per subject
By prosthetic

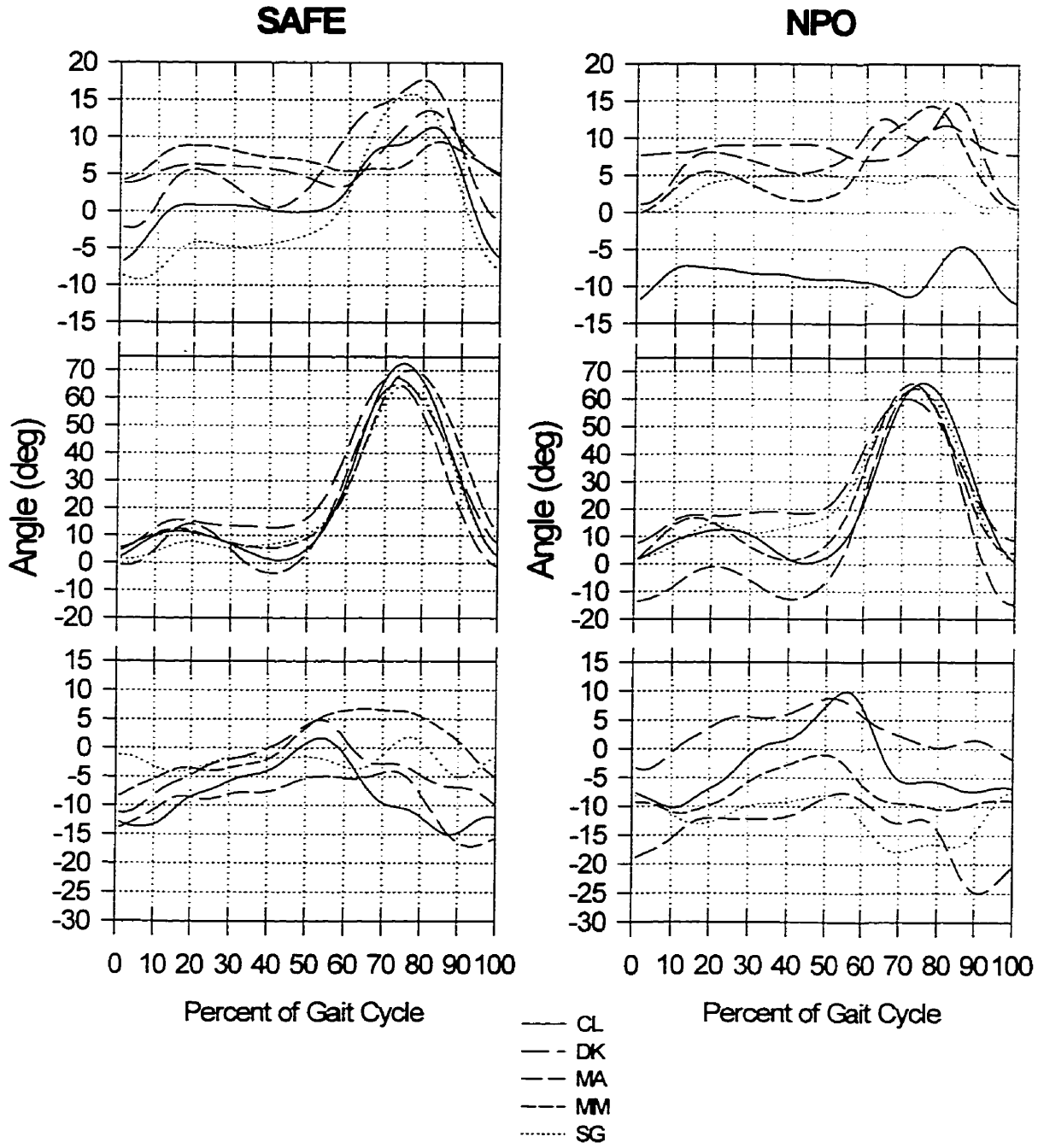


Figure G11: Average Relative Angle at the Knee Joint - Residual Limb
Five subjects
Average for five trials per subject
By prosthetic

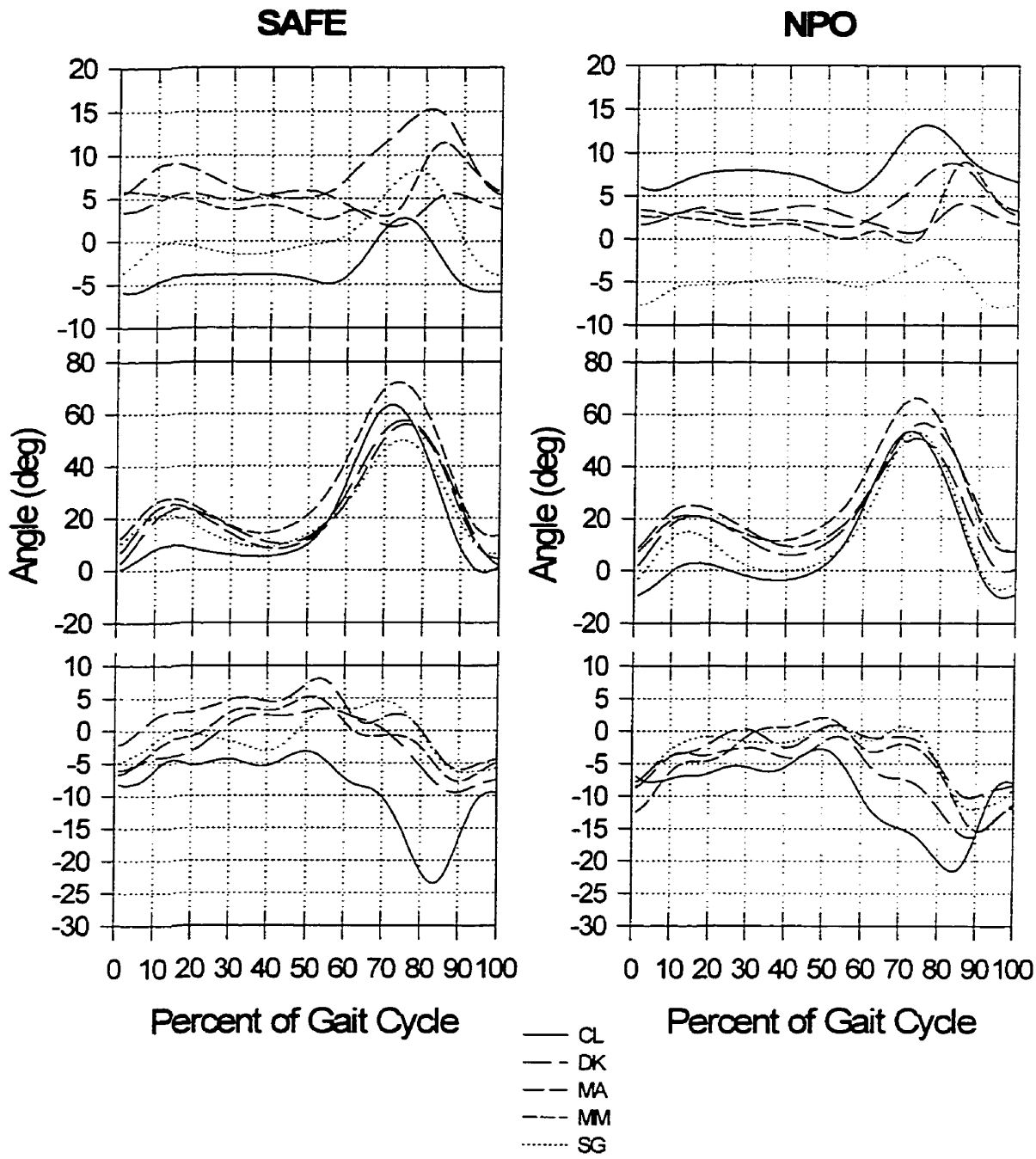


Figure G12: Average Relative Angle at the Knee Joint - Sound Limb
Five subjects
Average for five trials per subject
By prosthetic

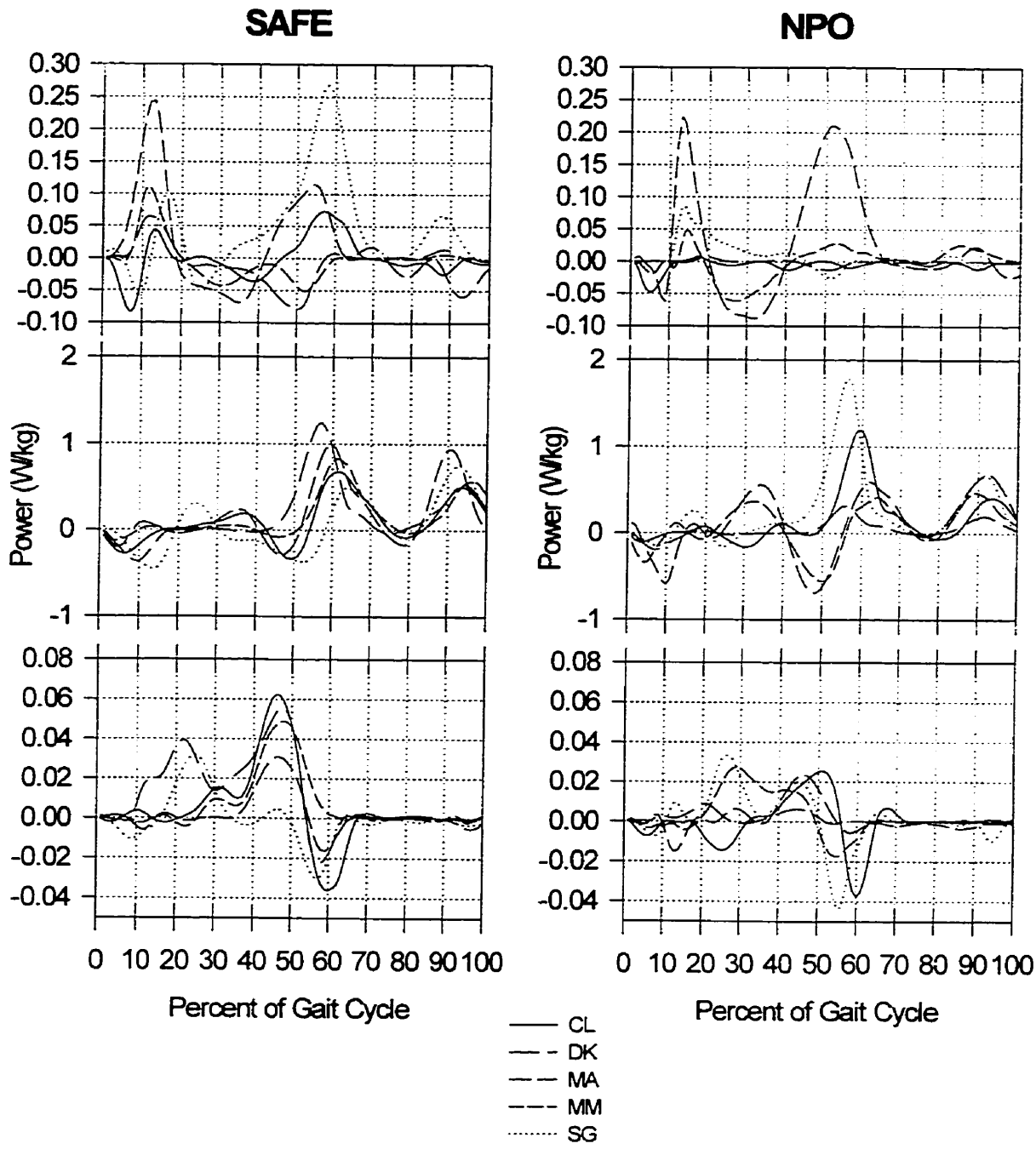


Figure G13: Average Power at the Knee Joint - Residual Limb
Five subjects
Average for five trials per subject
By prosthetic

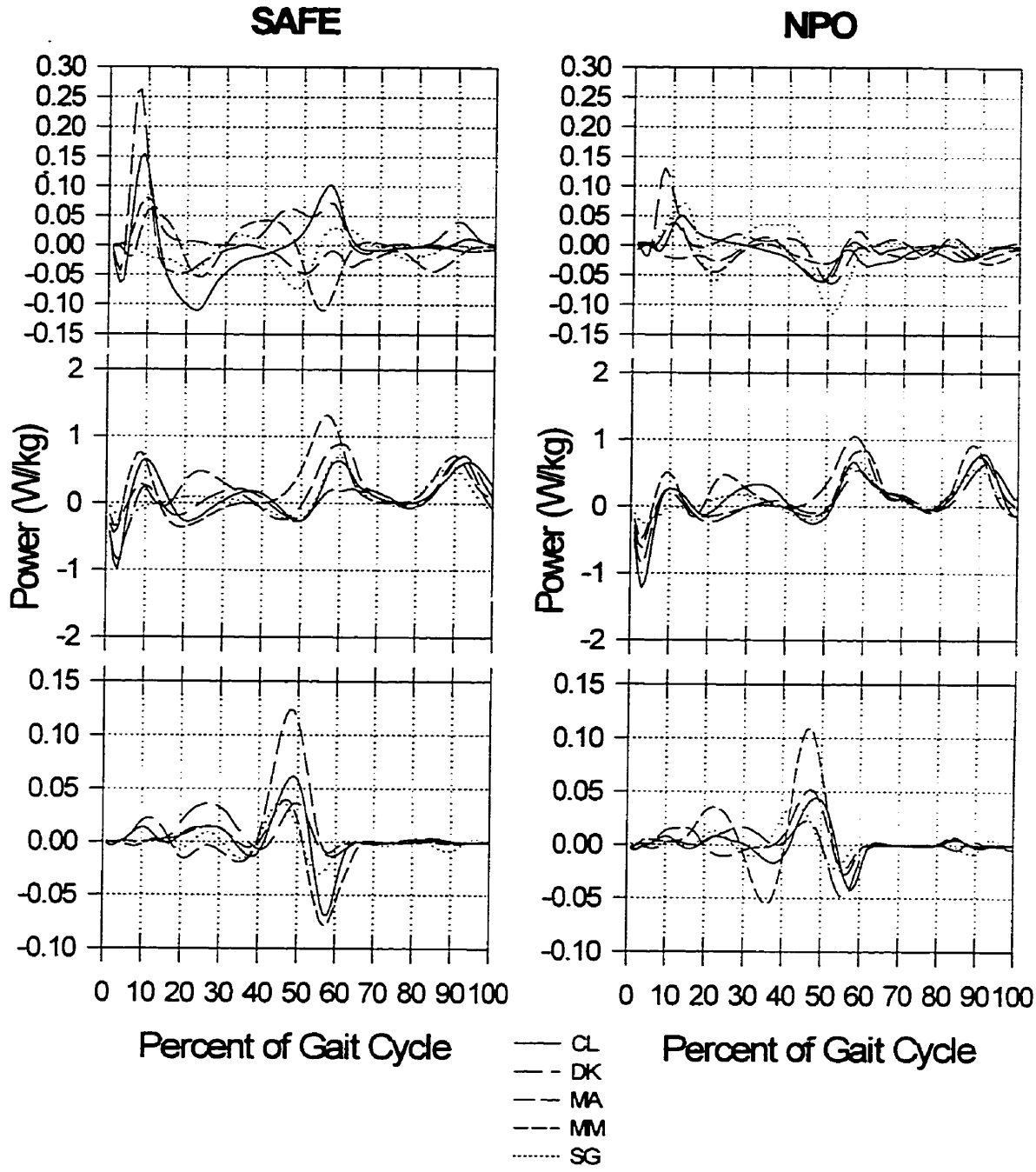


Figure G14: Average Power at the Knee Joint - Sound Limb
Five subjects
Average for five trials per subject
By prosthetic

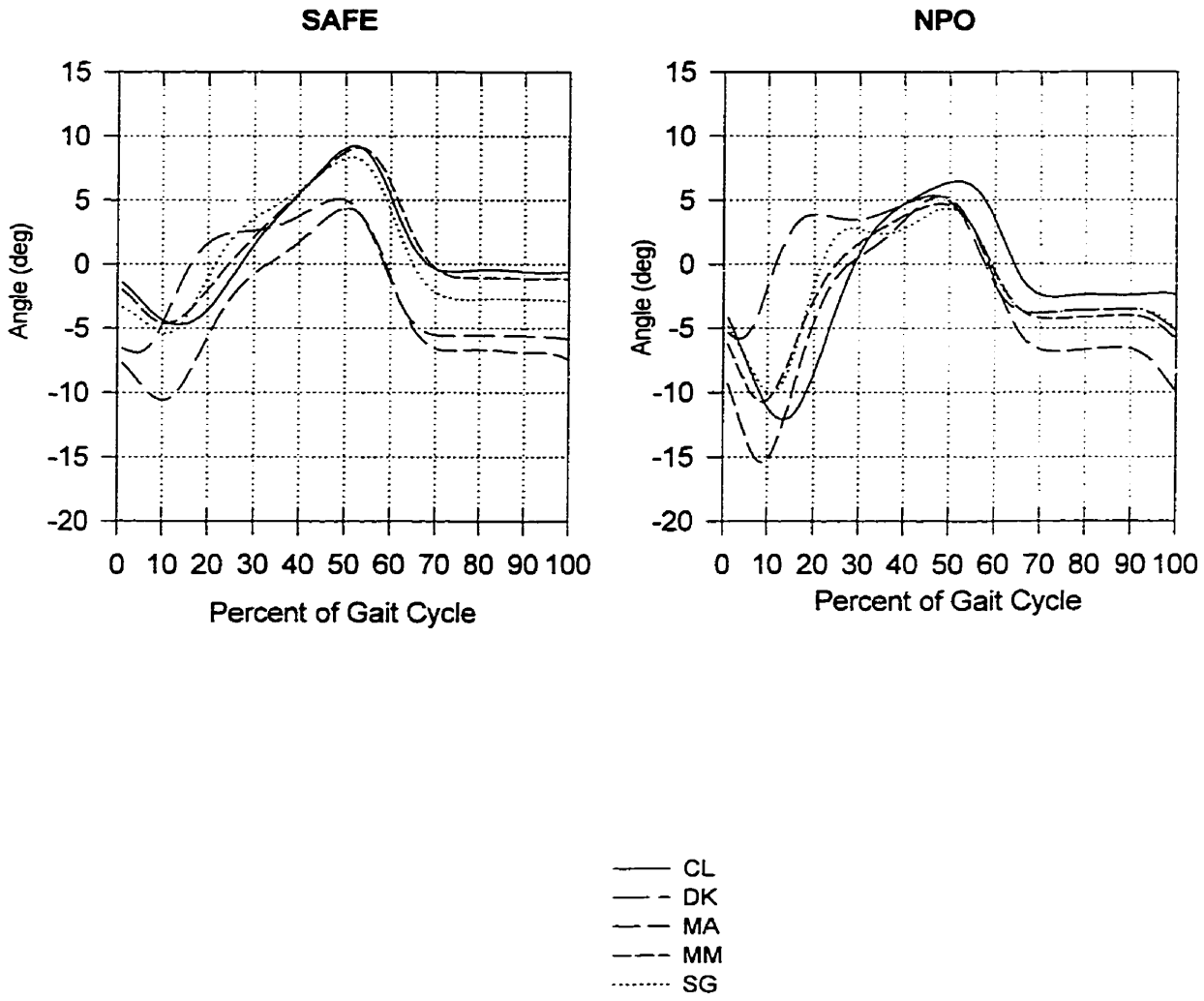


Figure G15: Average Relative Angle at the Ankle Joint - Residual Limb
Five subjects
Average for five trials per subject
By prosthetic

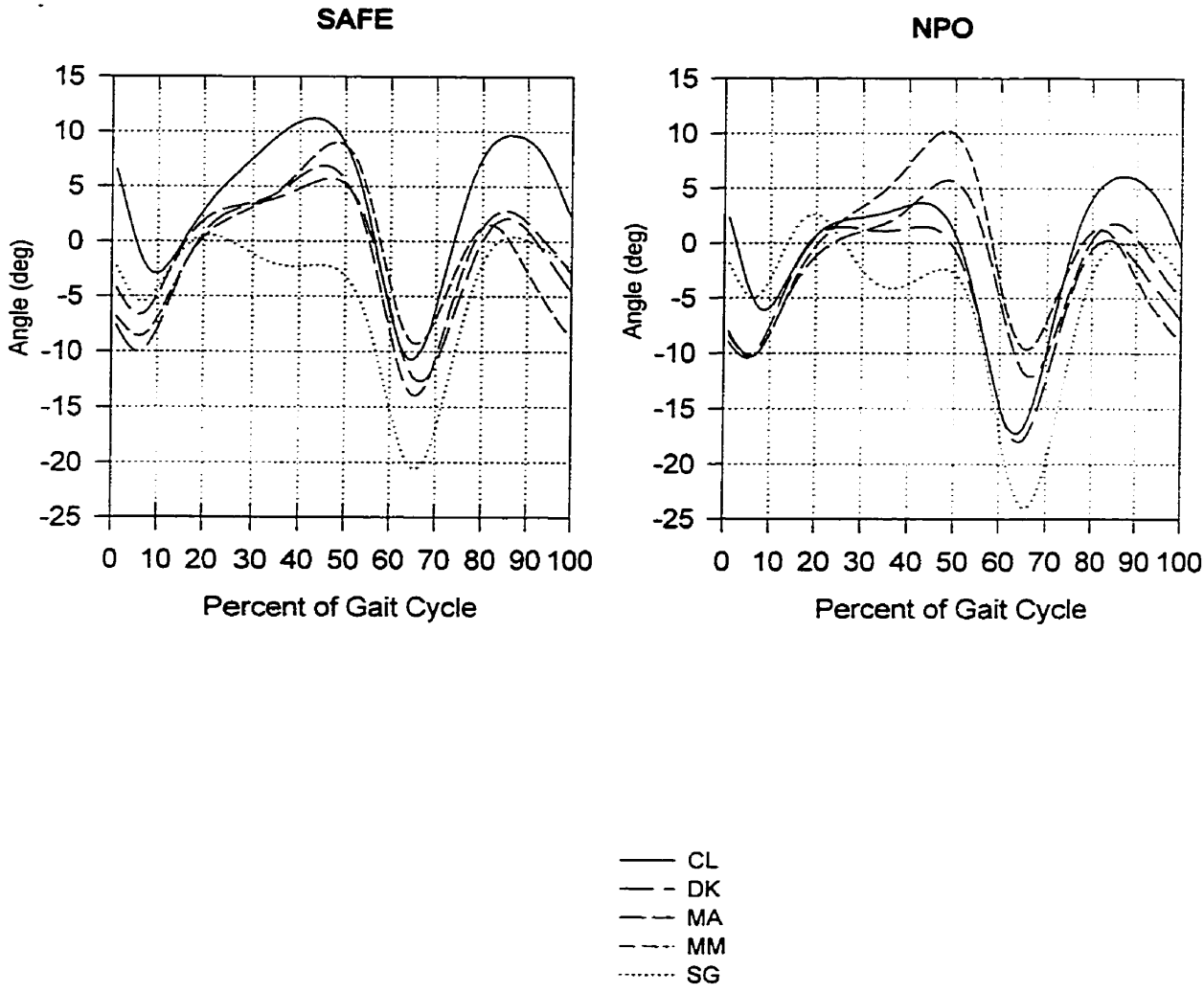


Figure G16: Average Relative Angle at the Ankle Joint - Sound Limb
Five subjects
Average for five trials per subject
By prosthetic

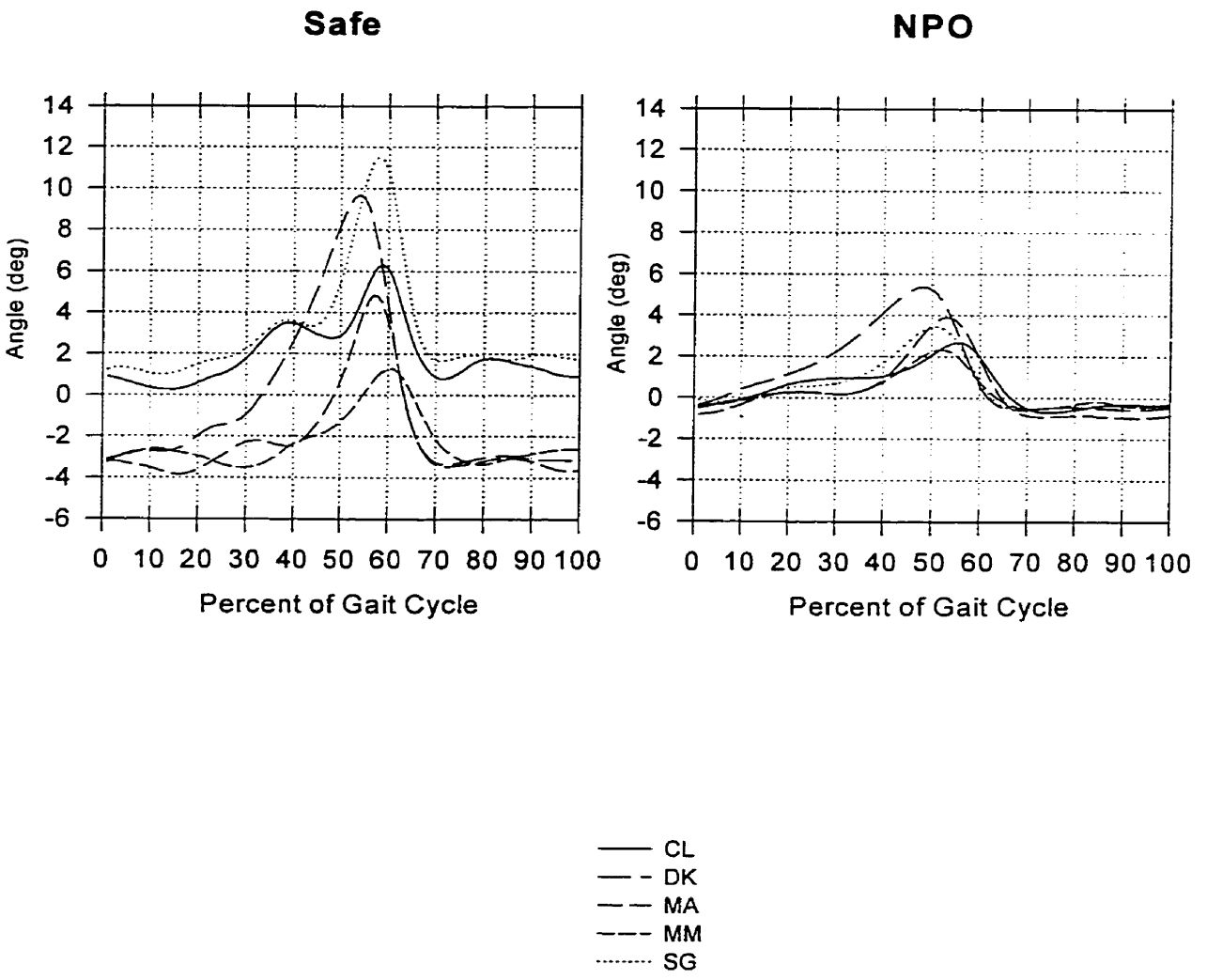


Figure G18: Average Relative Angle at the 5th Metatarsal Joint - Sound Limb
Five subjects
Average for five trials per subject
By prosthetic

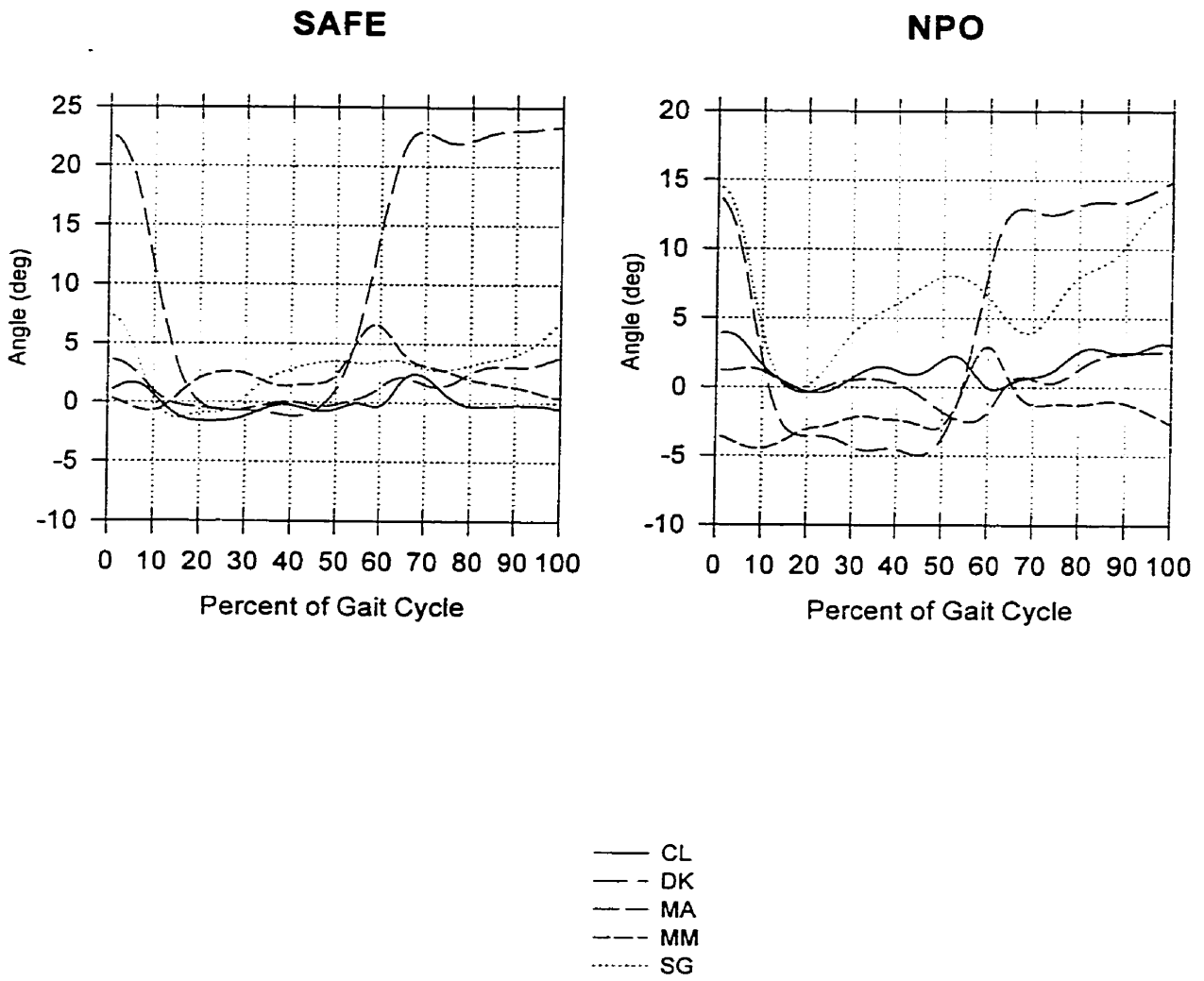


Figure G17: Average Relative Angle at the 5th Metatarsal Joint - Residual Limb
Five subjects
Average for five trials per subject
By prosthetic

Appendix H

Curve Parameters

Appendix H
Curve Parameters
H-2

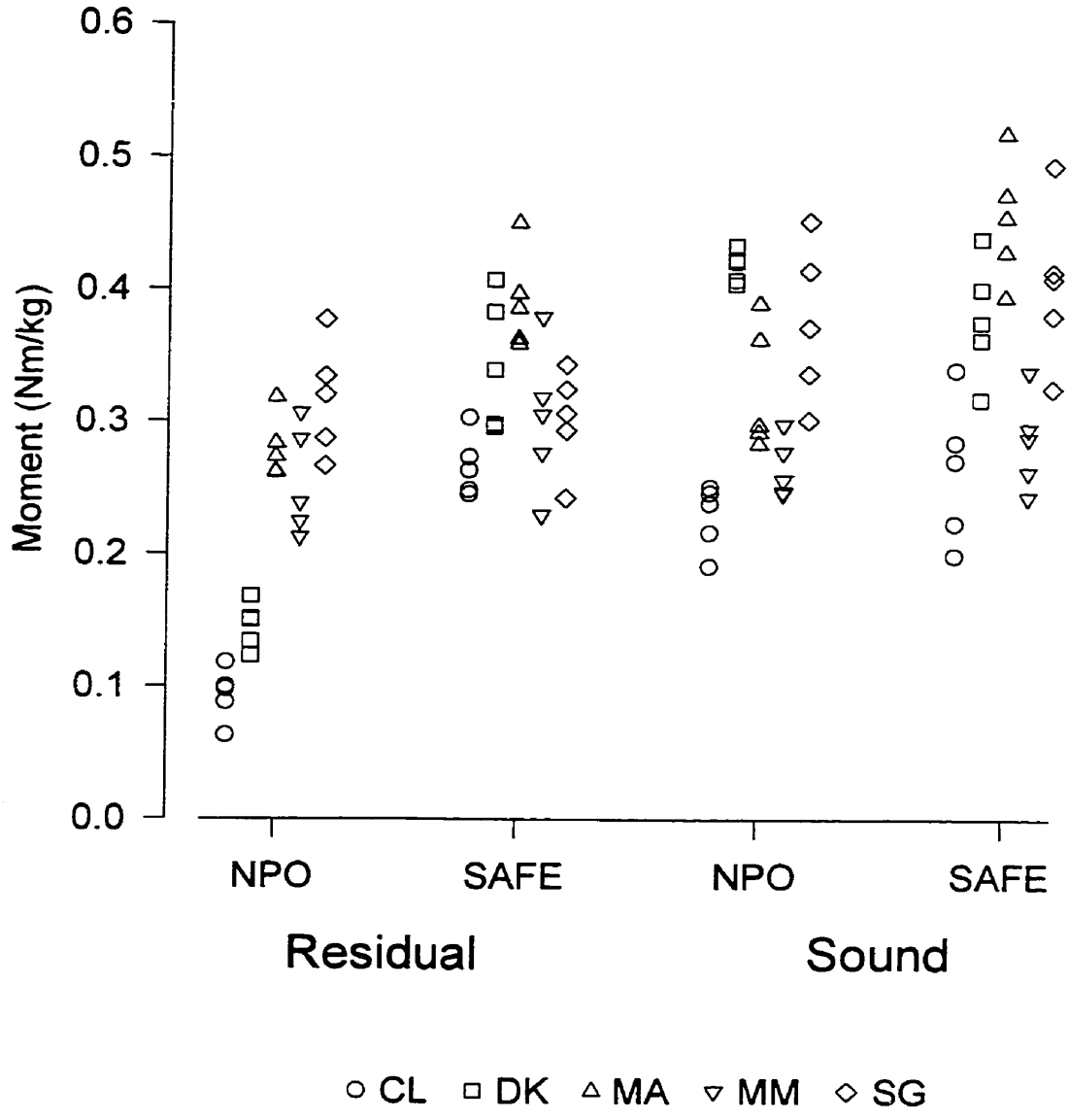


Figure H1: Maximum Moment PA Axis (0-40% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

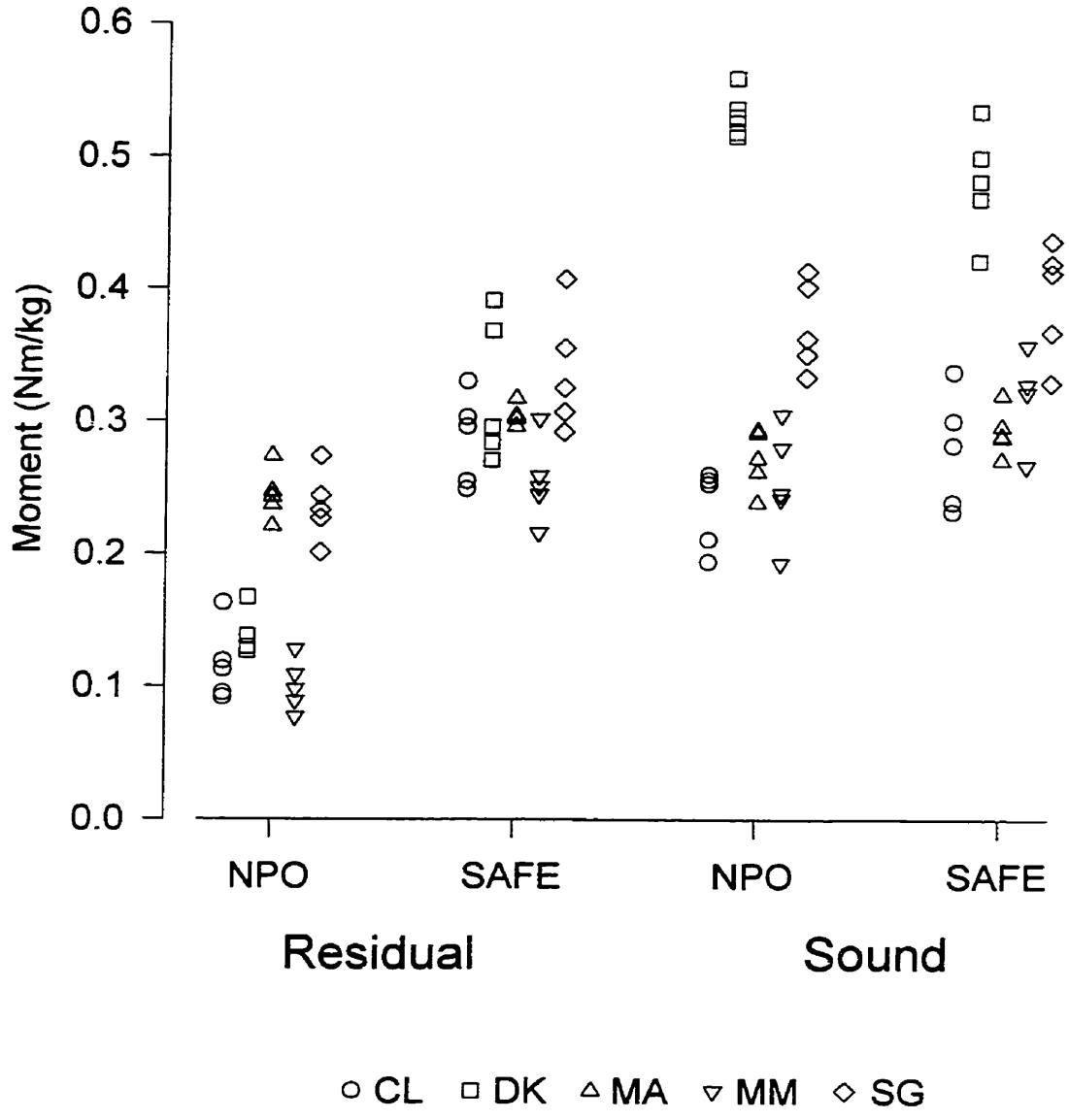


Figure H2: Maximum Moment PA Axis (40-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

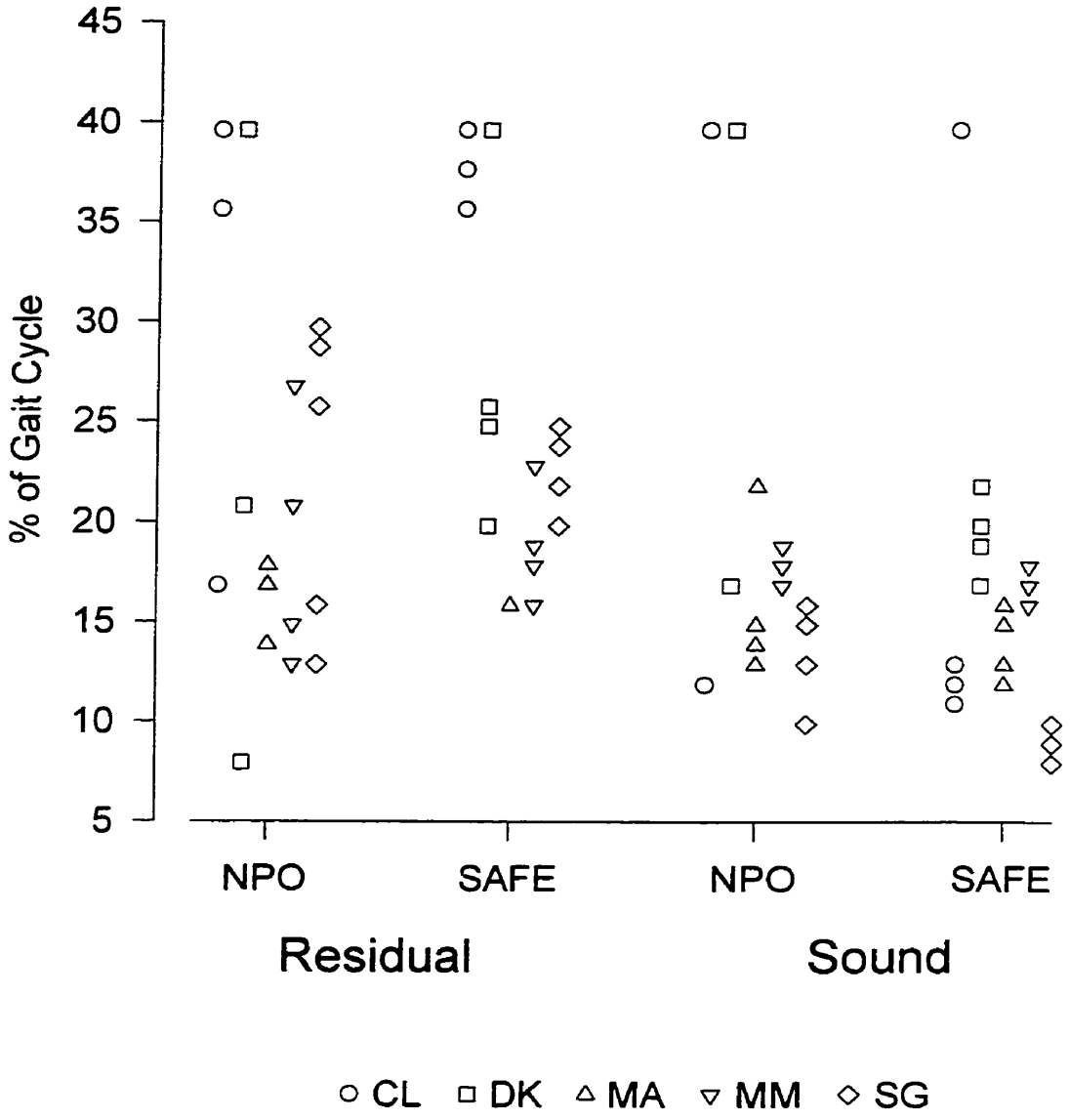


Figure H3: Location of Maximum Moment PA Axis (0-40% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

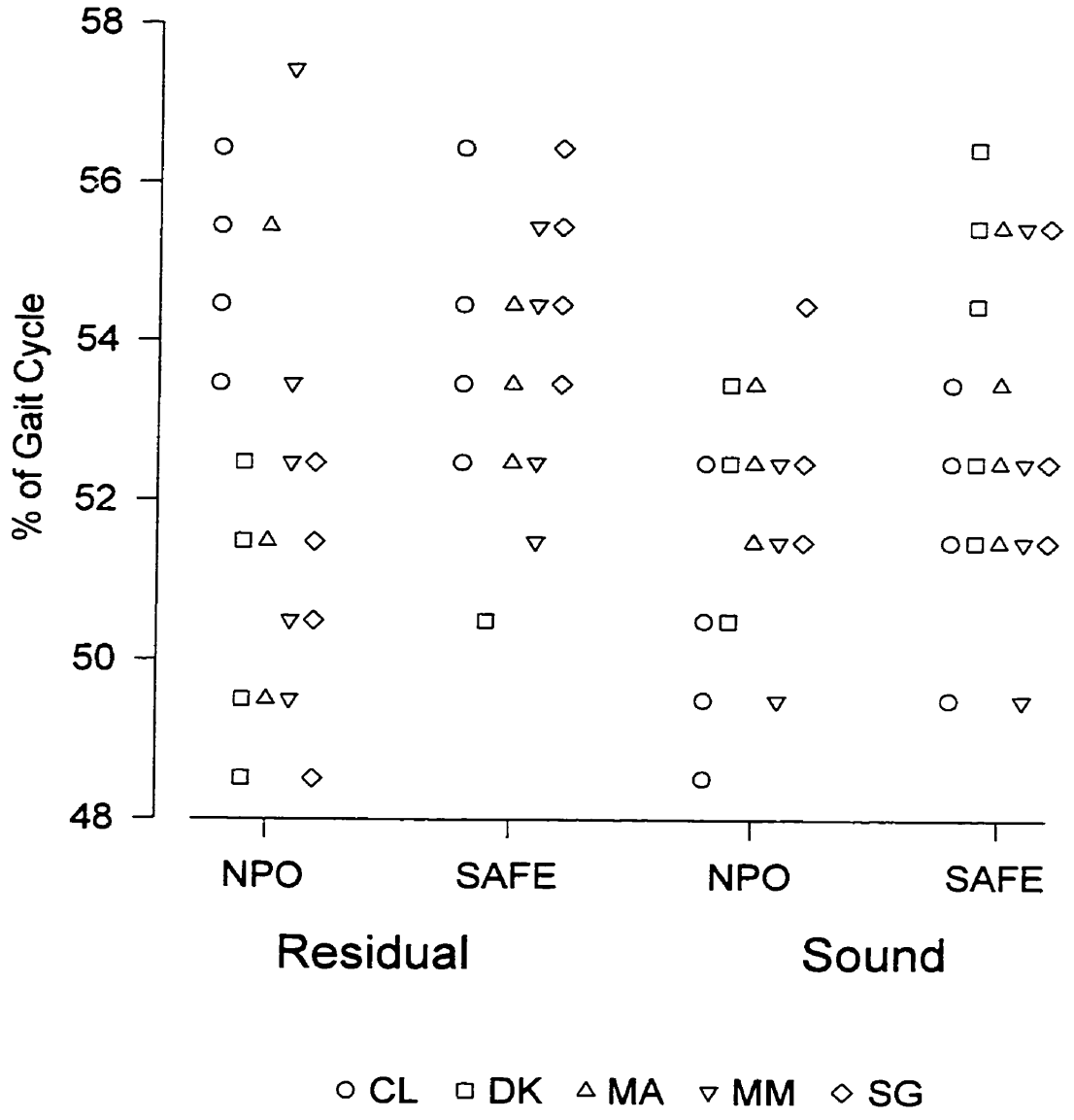


Figure H4: Location of Maximum Moment PA Axis (40-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

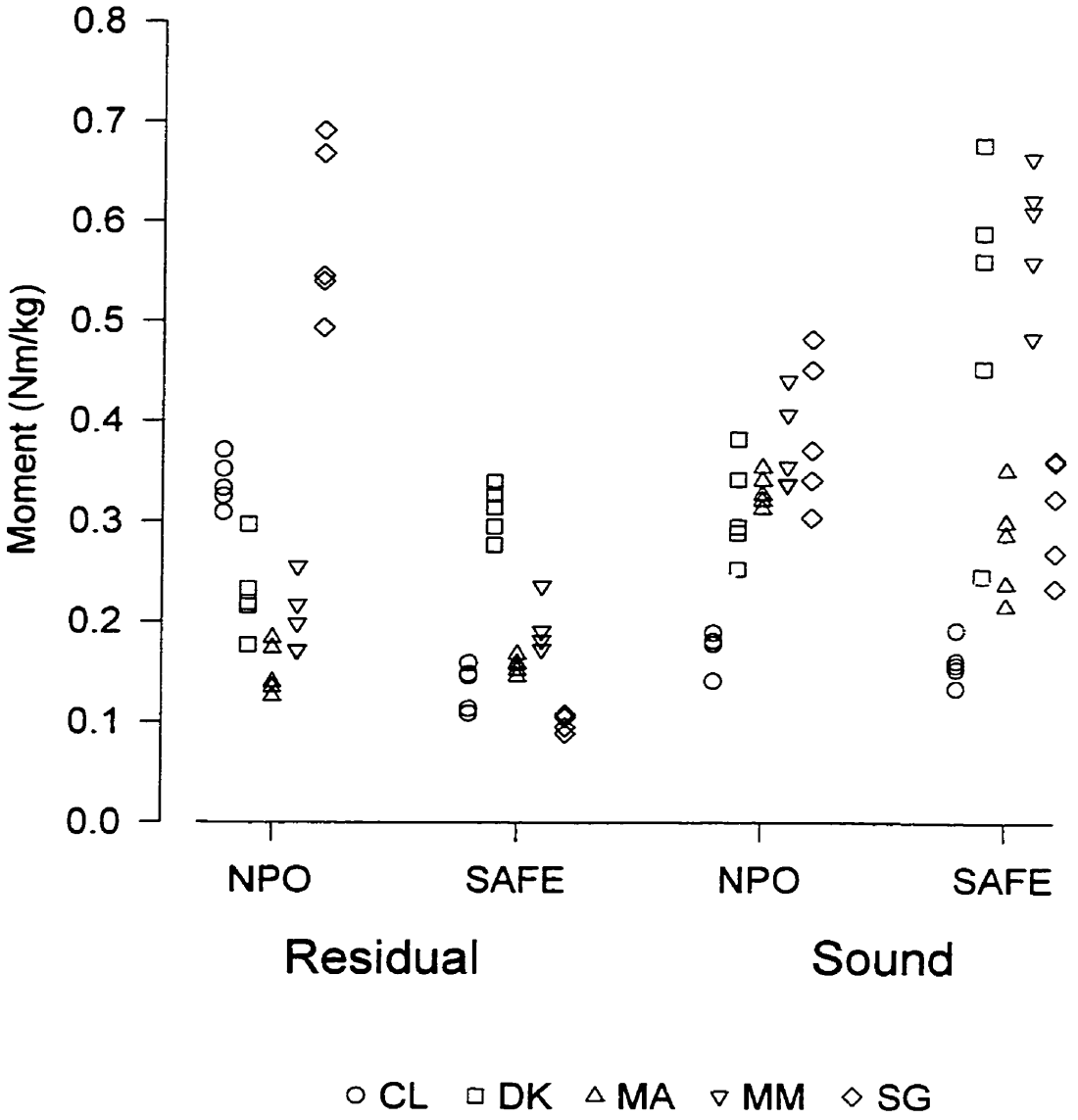


Figure H5: Maximum Moment LM Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

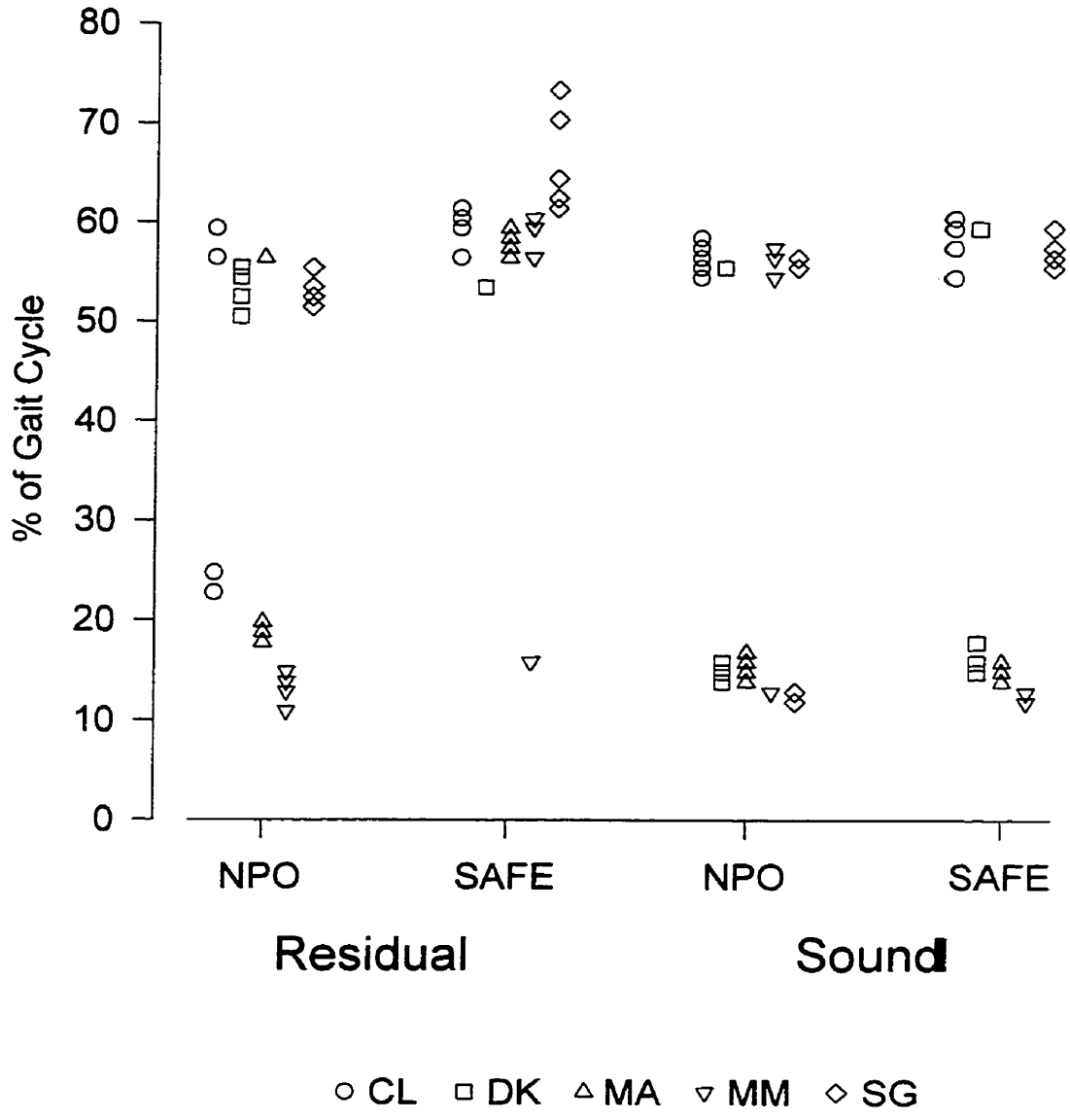


Figure H6: Location of Maximum Moment LM Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

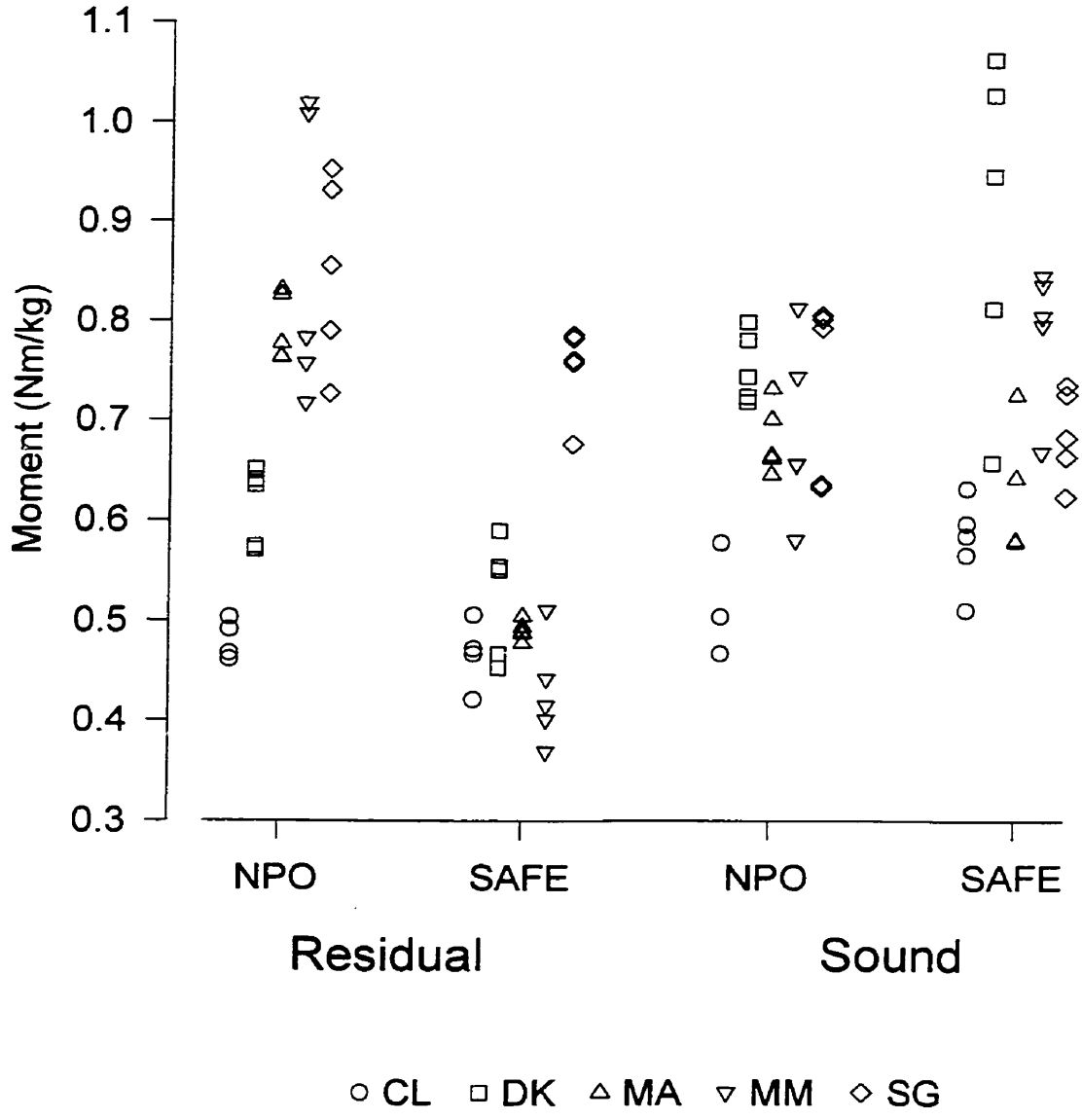


Figure H7: Range of Moment LM Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

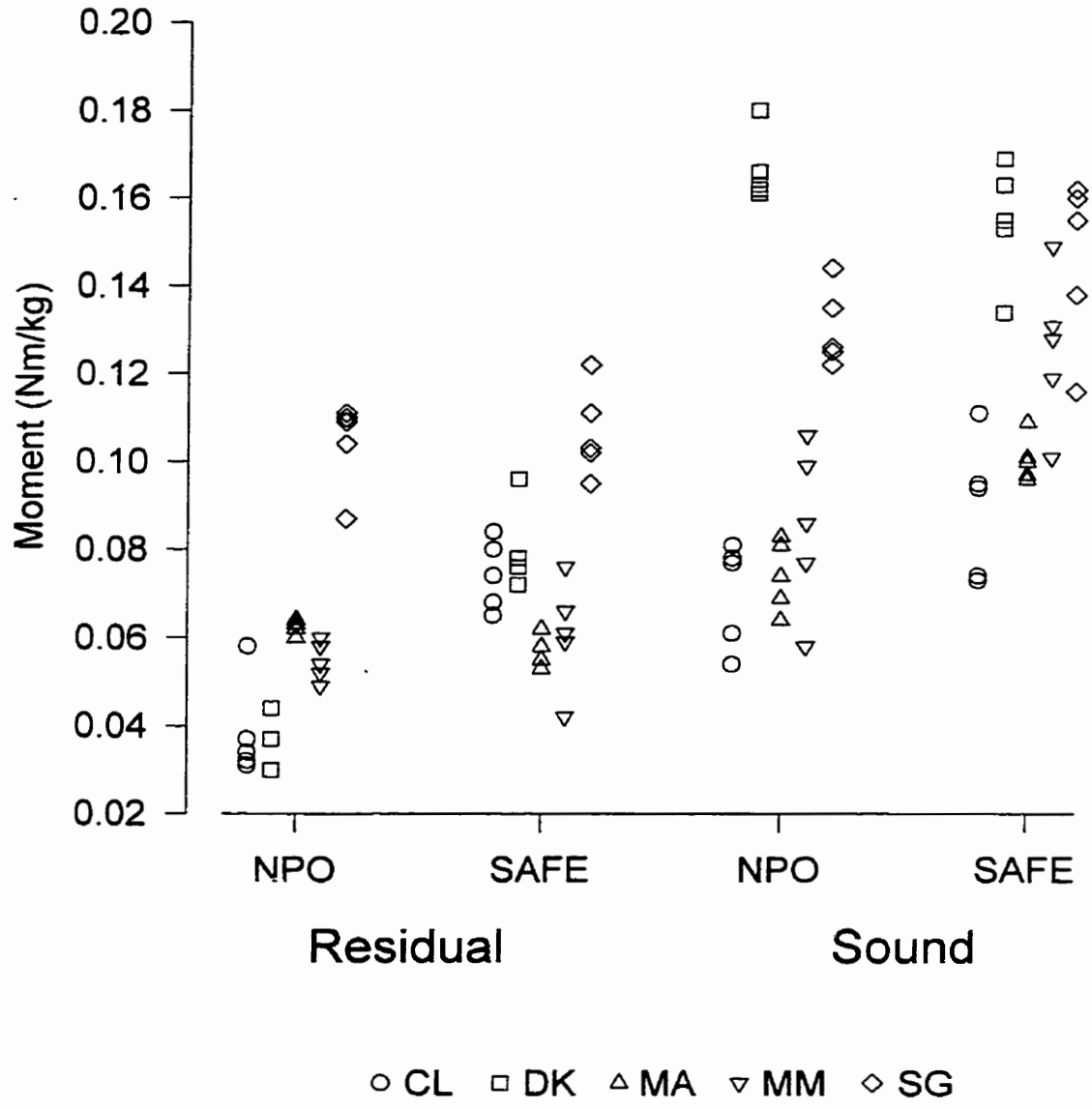


Figure H8: Maximum Moment DP Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

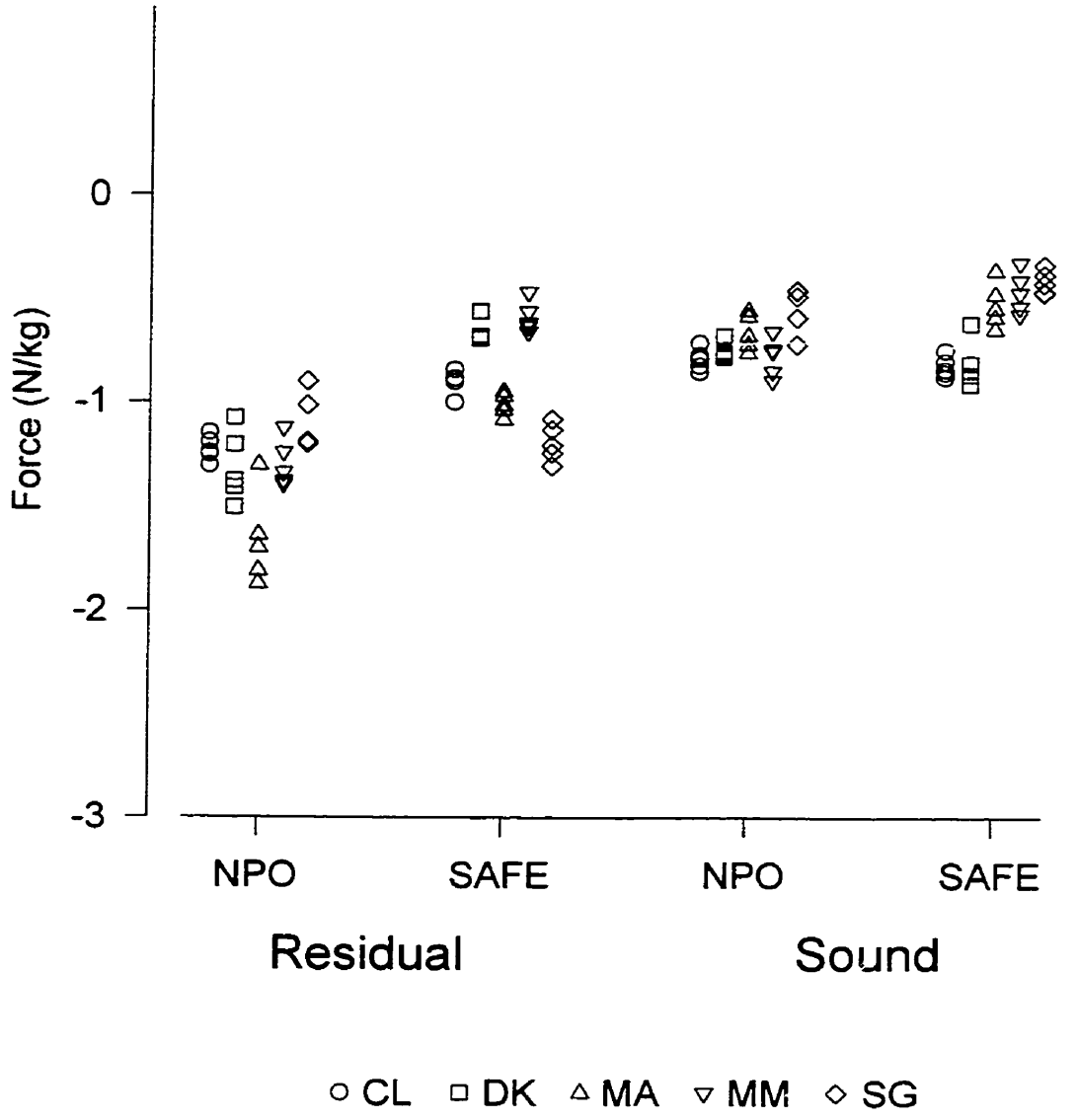


Figure H9: Minimum Force PA Axis (0-20% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

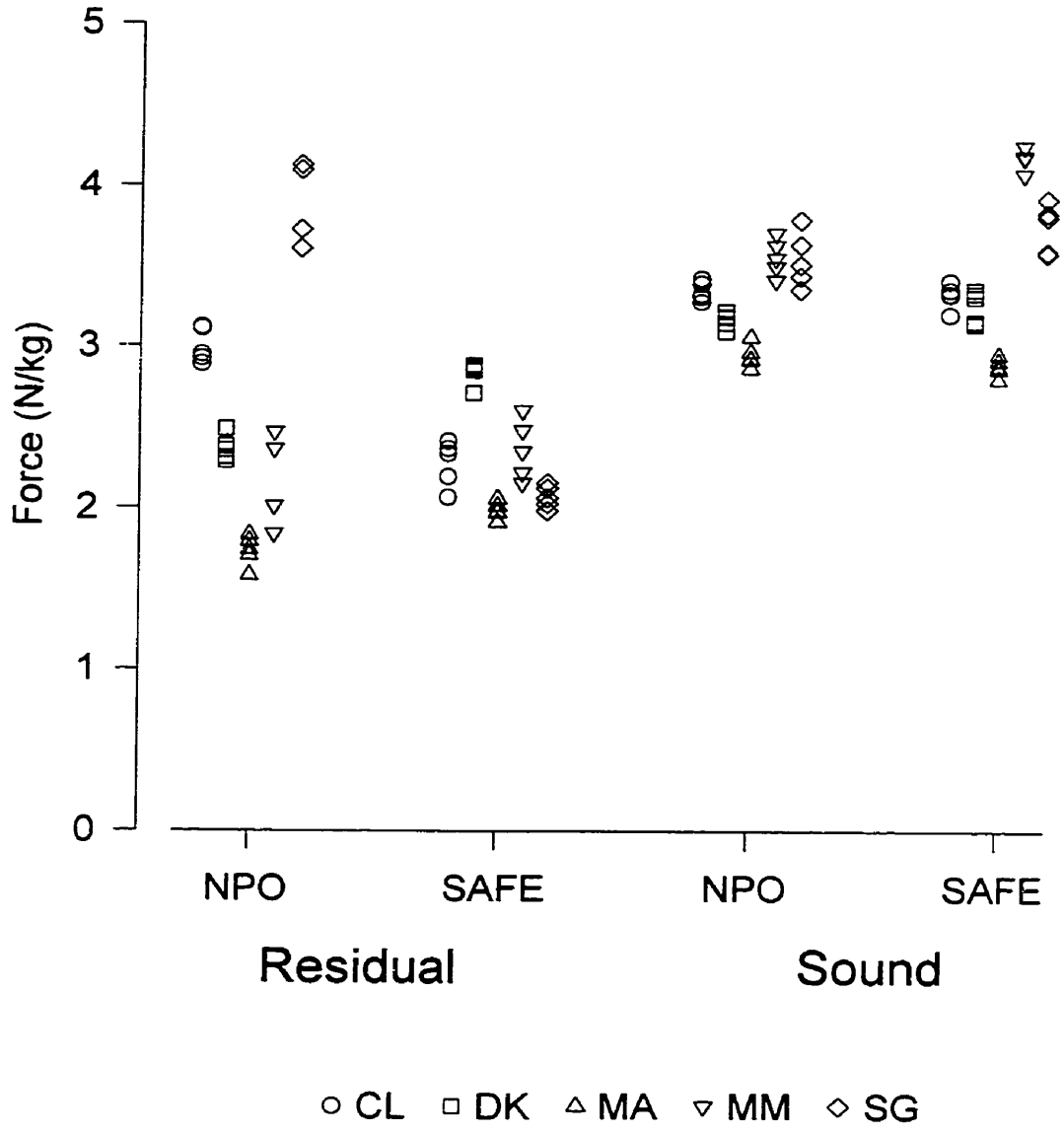


Figure H10: Maximum Force PA Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

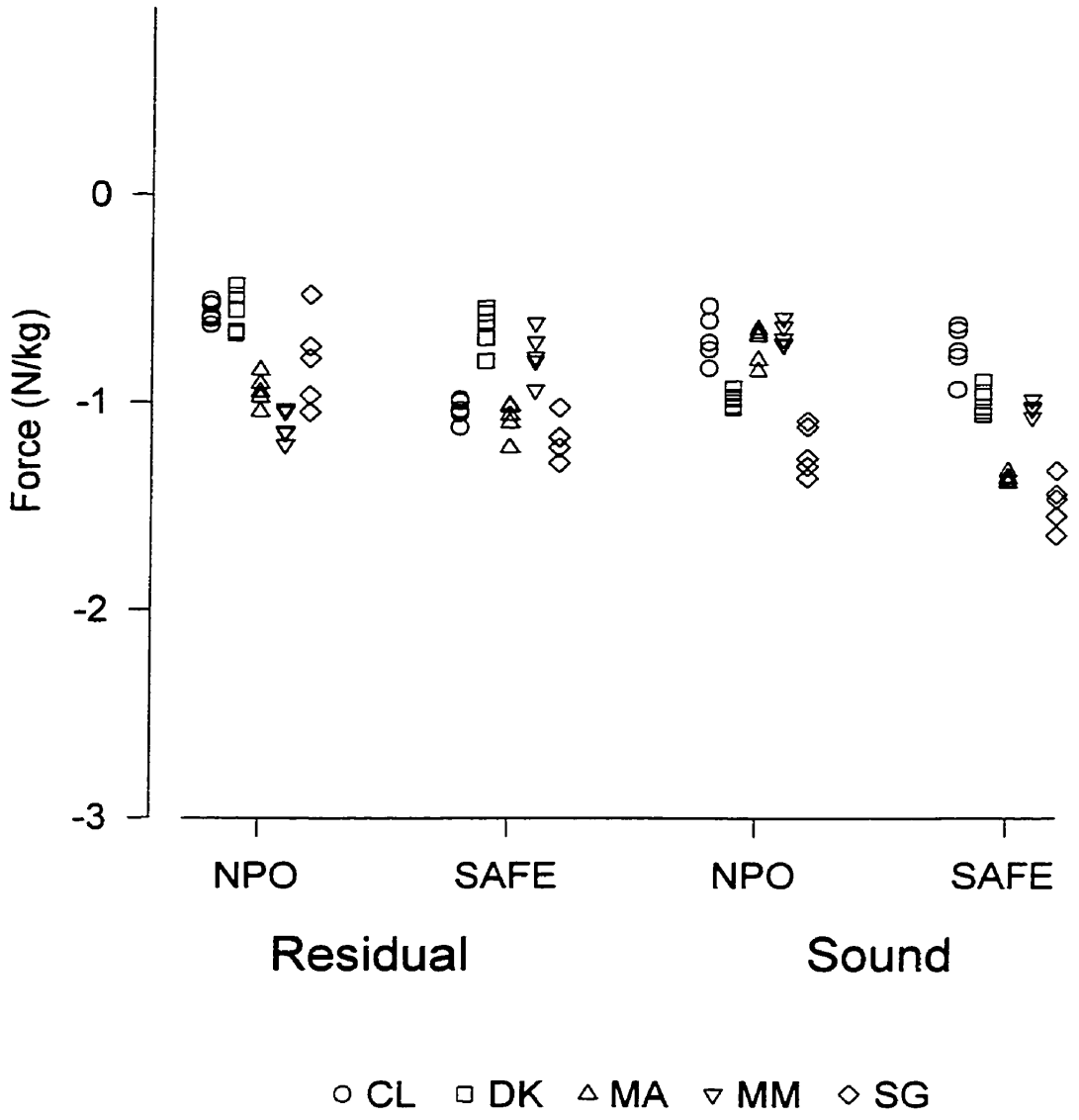


Figure H11: Minimum Force LM Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

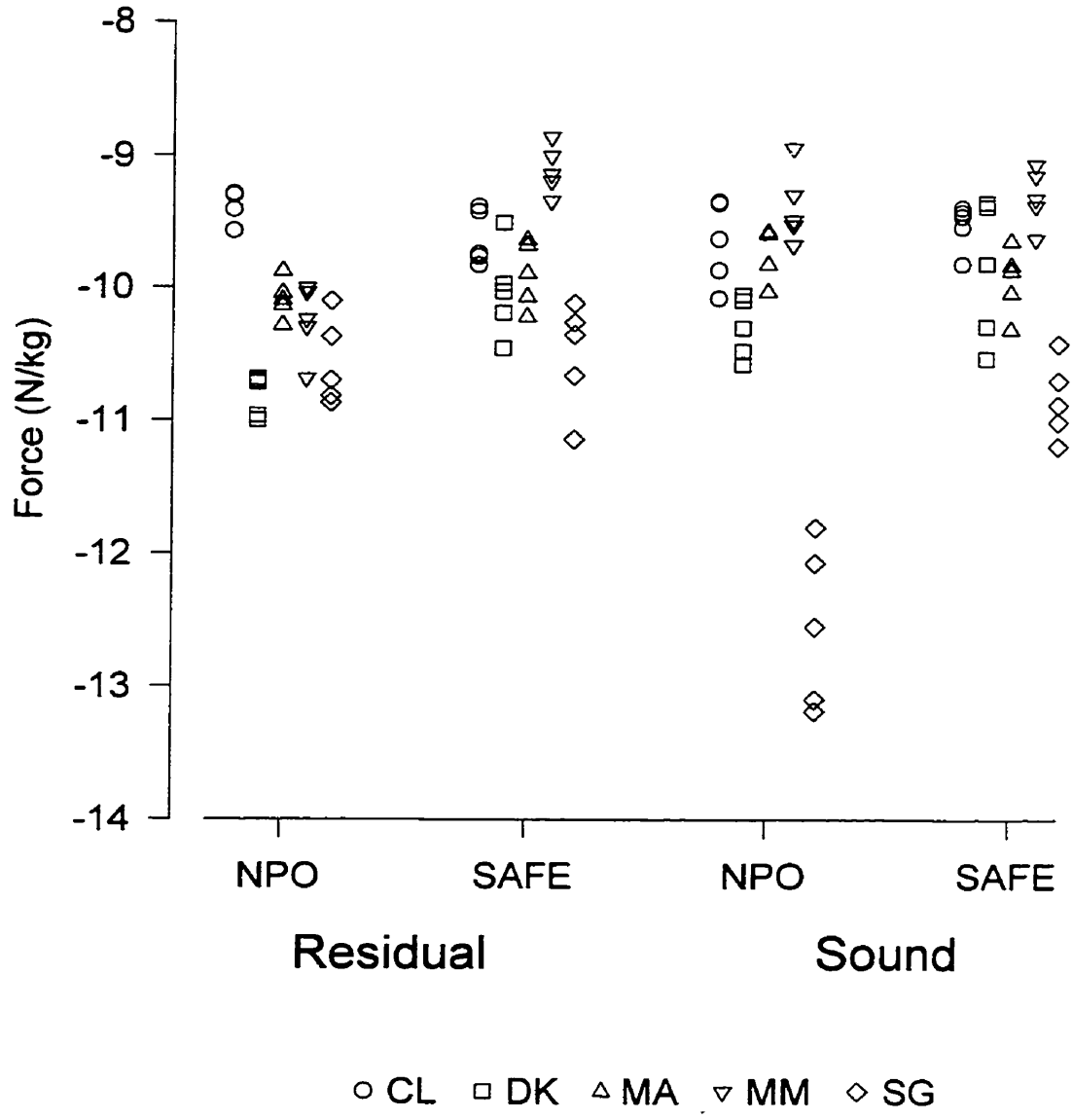


Figure H12: Minimum Force DP Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

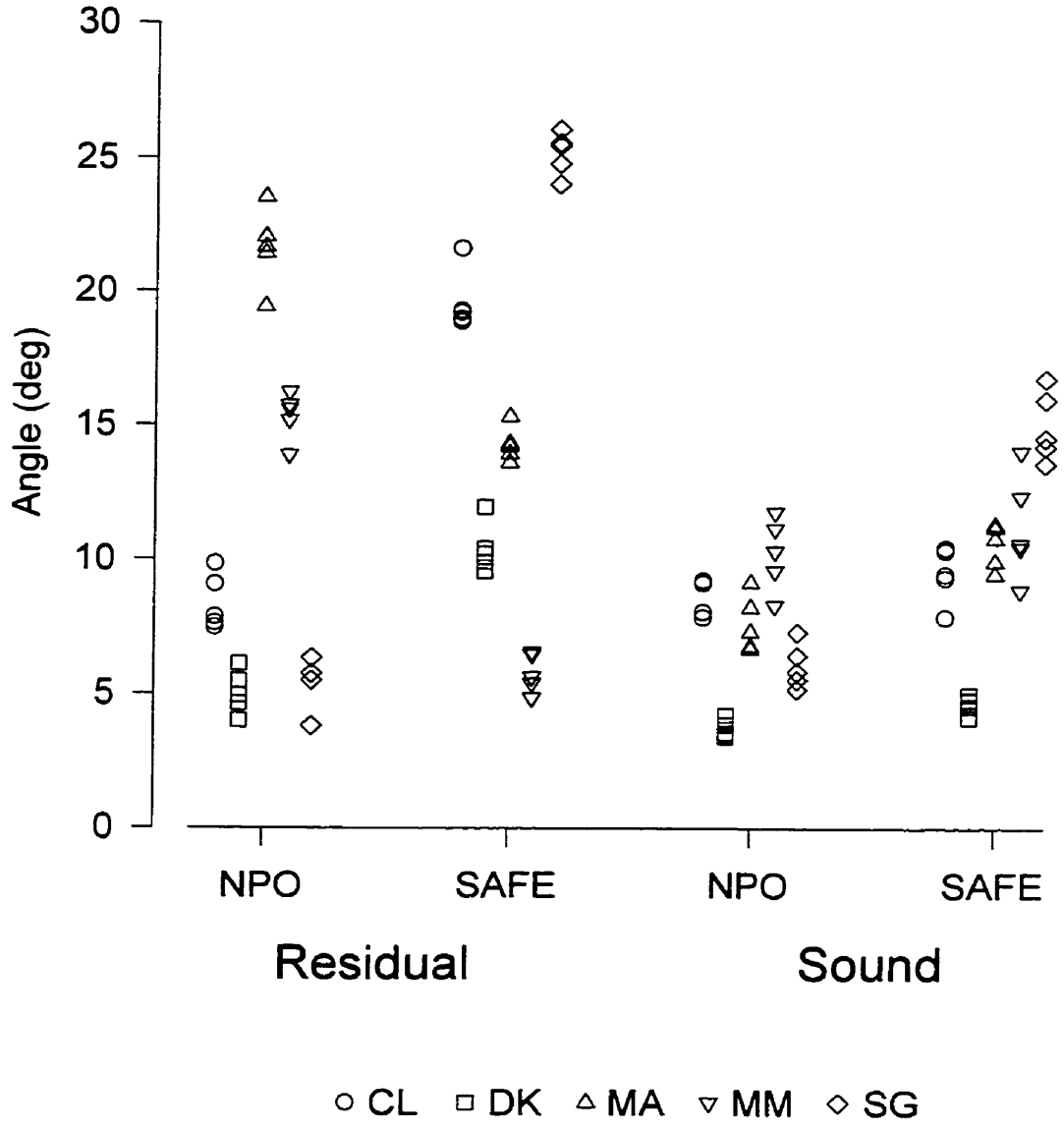


Figure H13: Range of Knee Angle PA Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

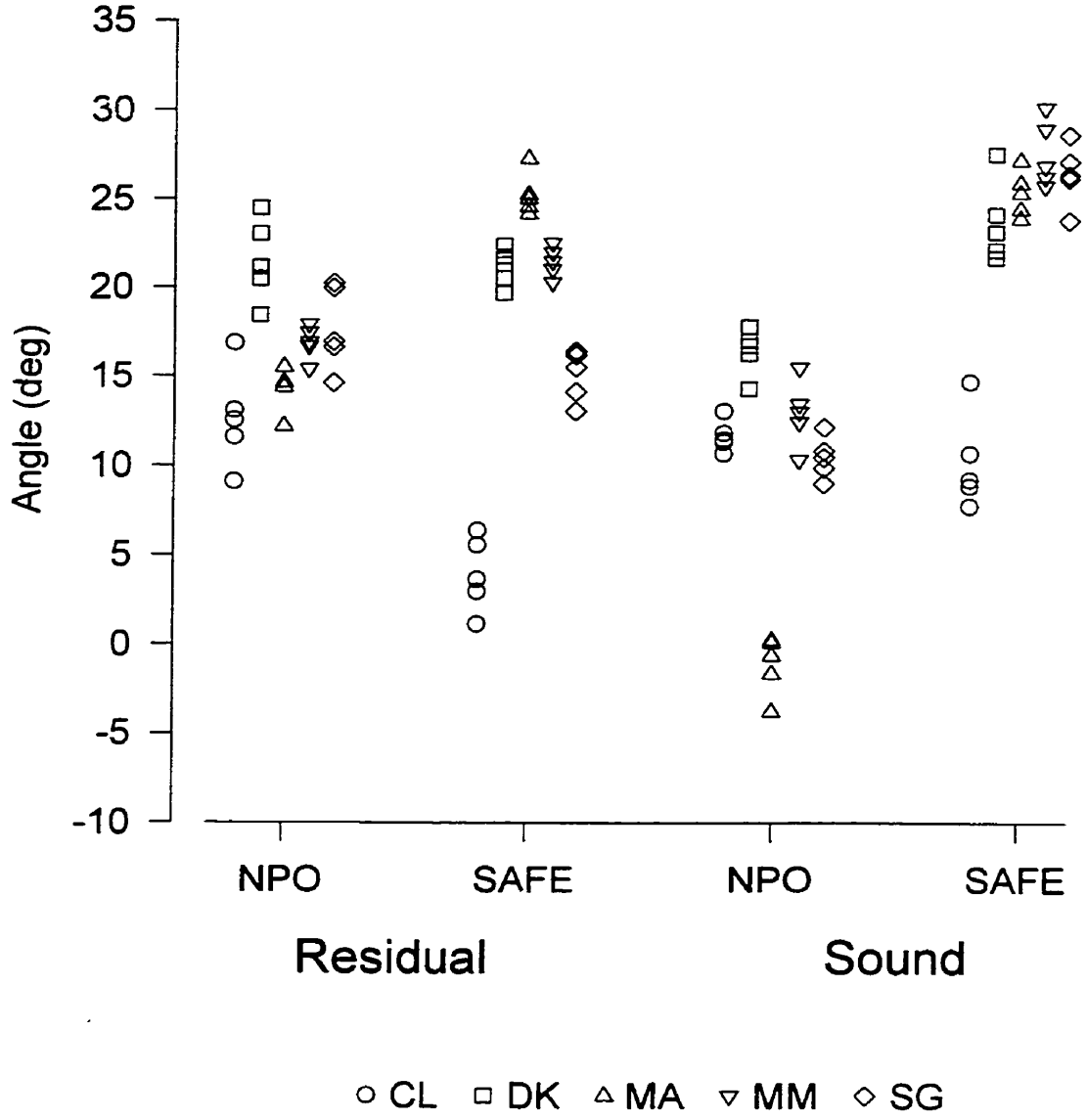


Figure H14: Maximum Angle LM Axis (0-50% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

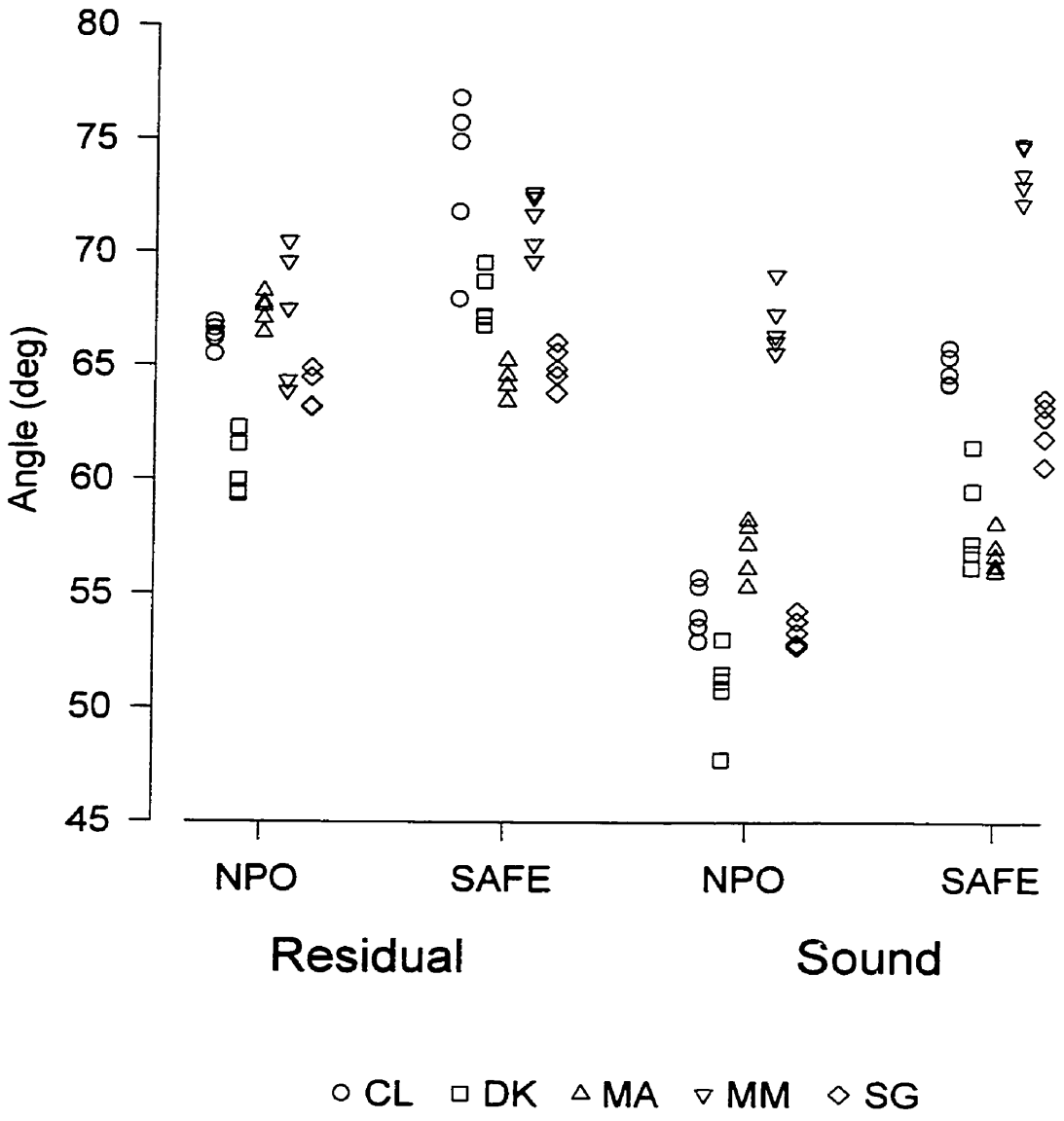


Figure H15: Maximum Angle LM Axis (50-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

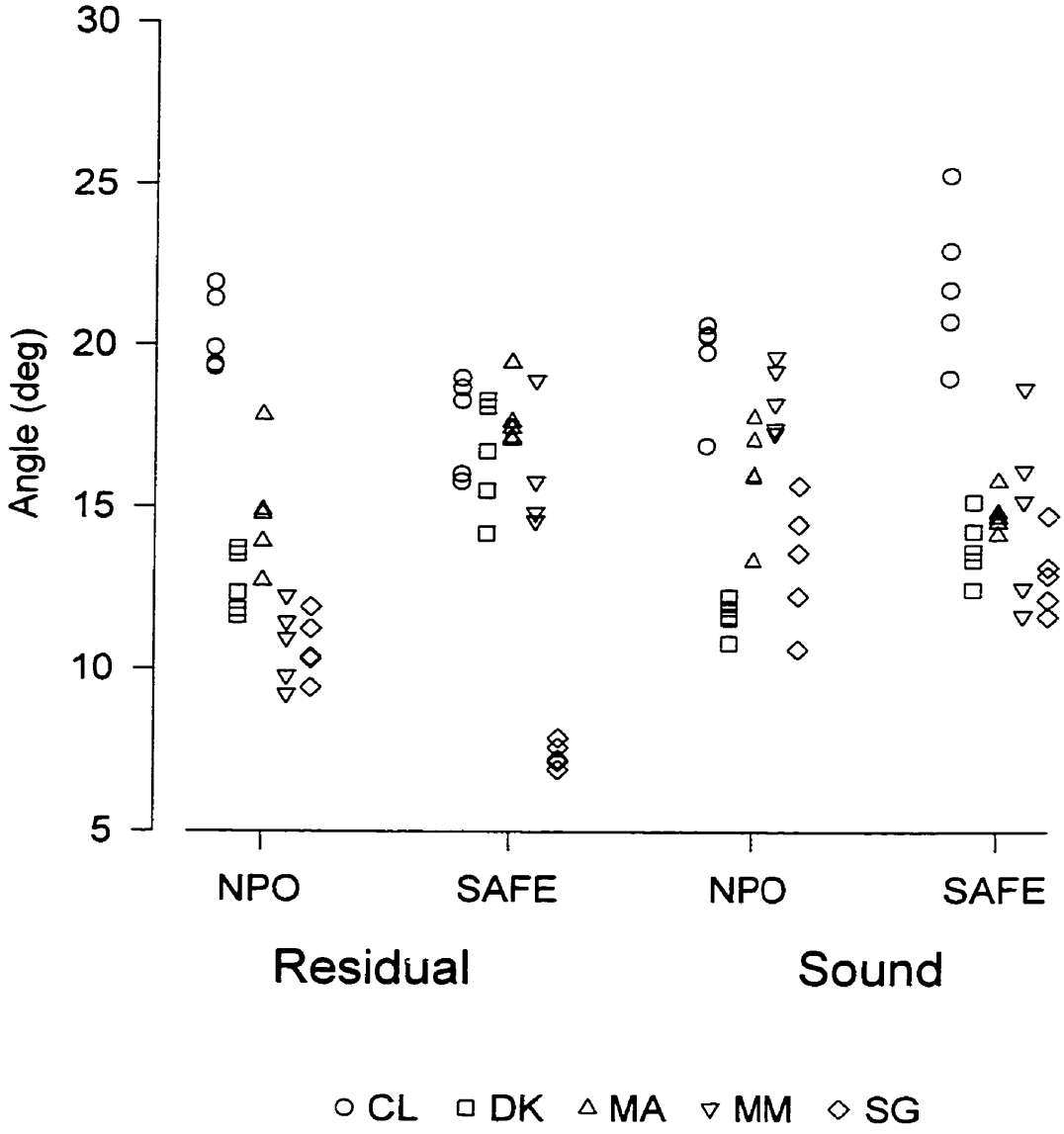


Figure H16: Range of Angle DP Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

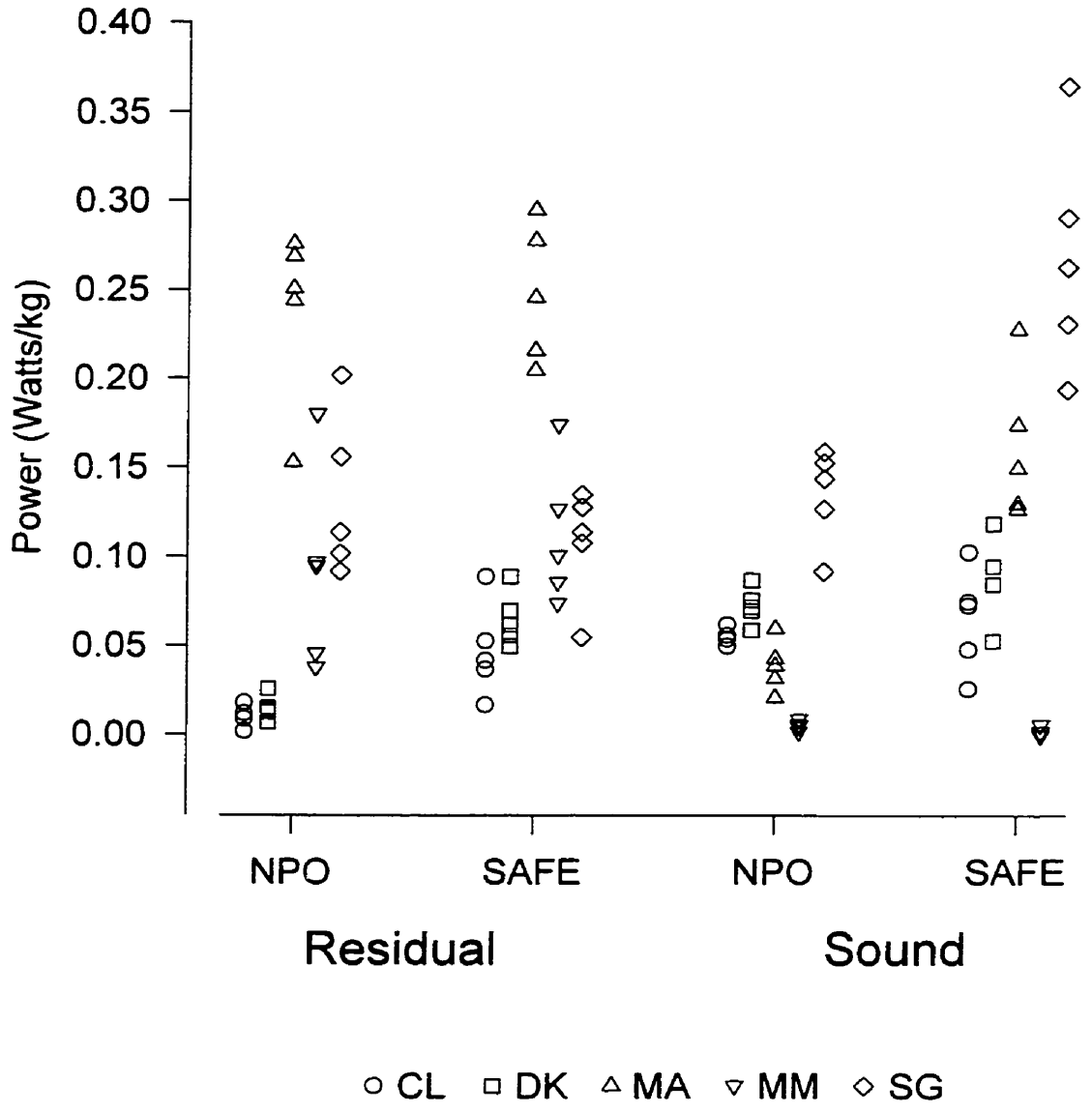


Figure H17: Maximum Power PA Axis (0-20% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

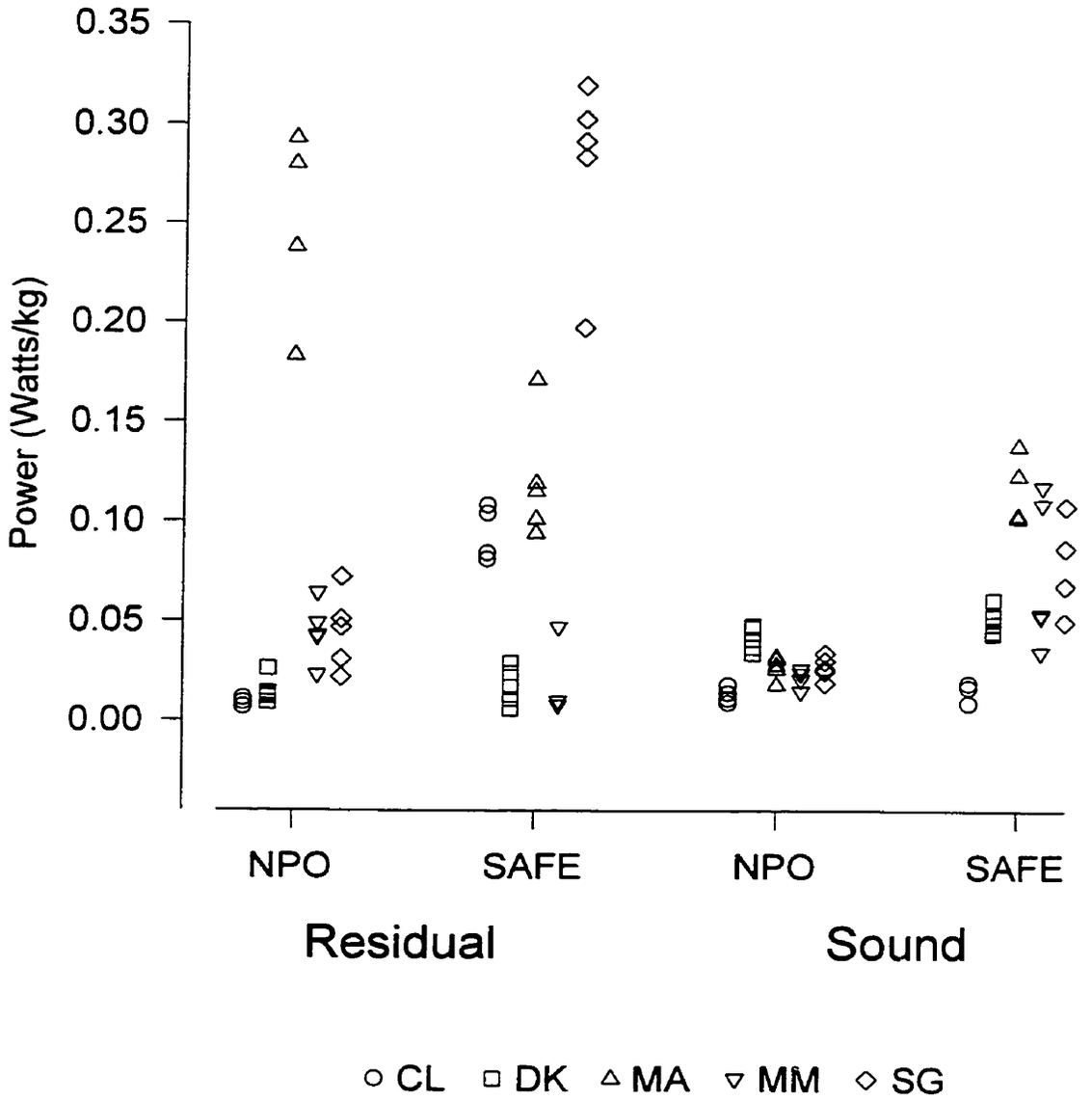


Figure H18: Maximum Power PA Axis (20-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

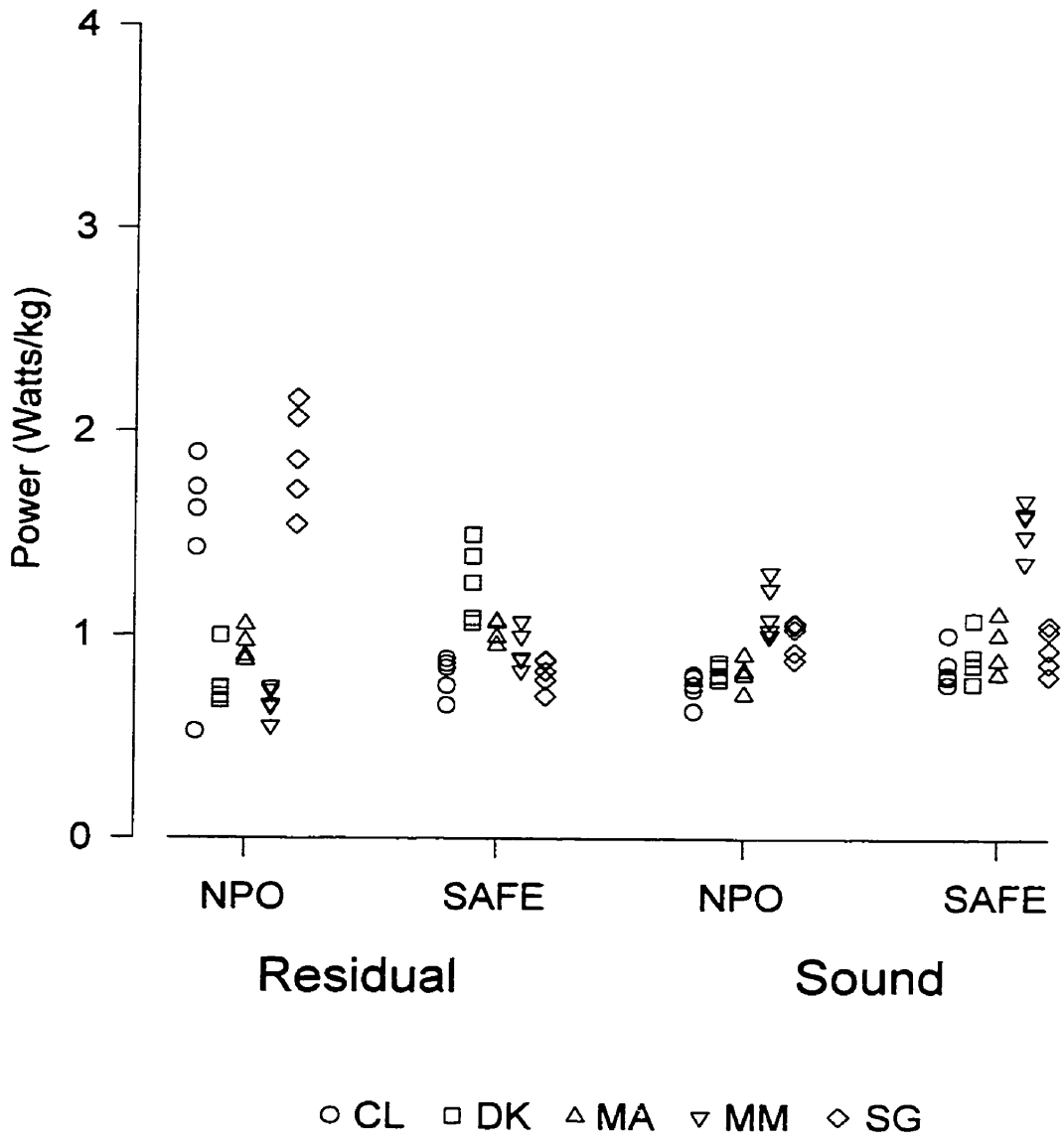


Figure H19: Maximum Power LM Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

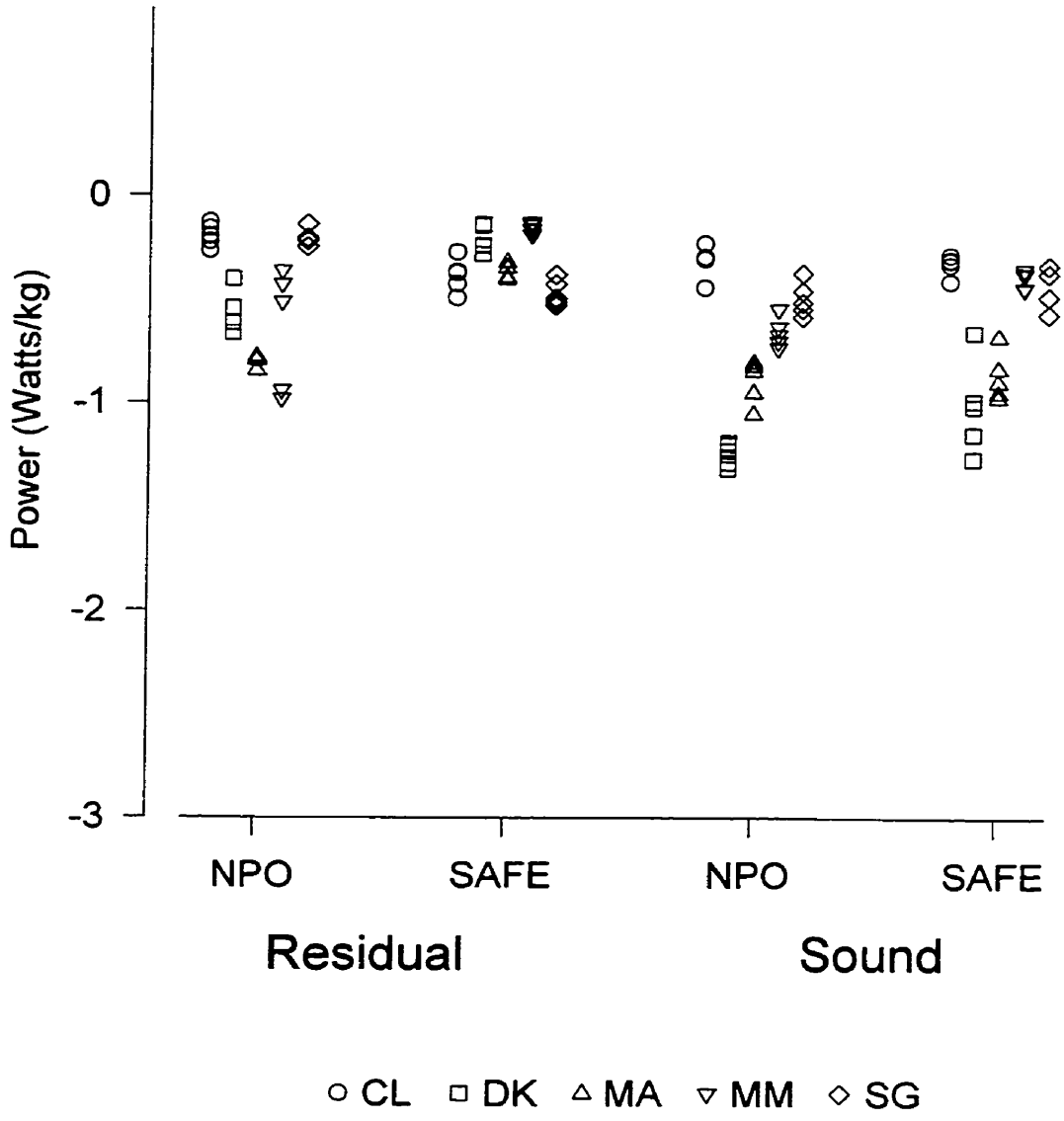


Figure H20: Minimum Power LM Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

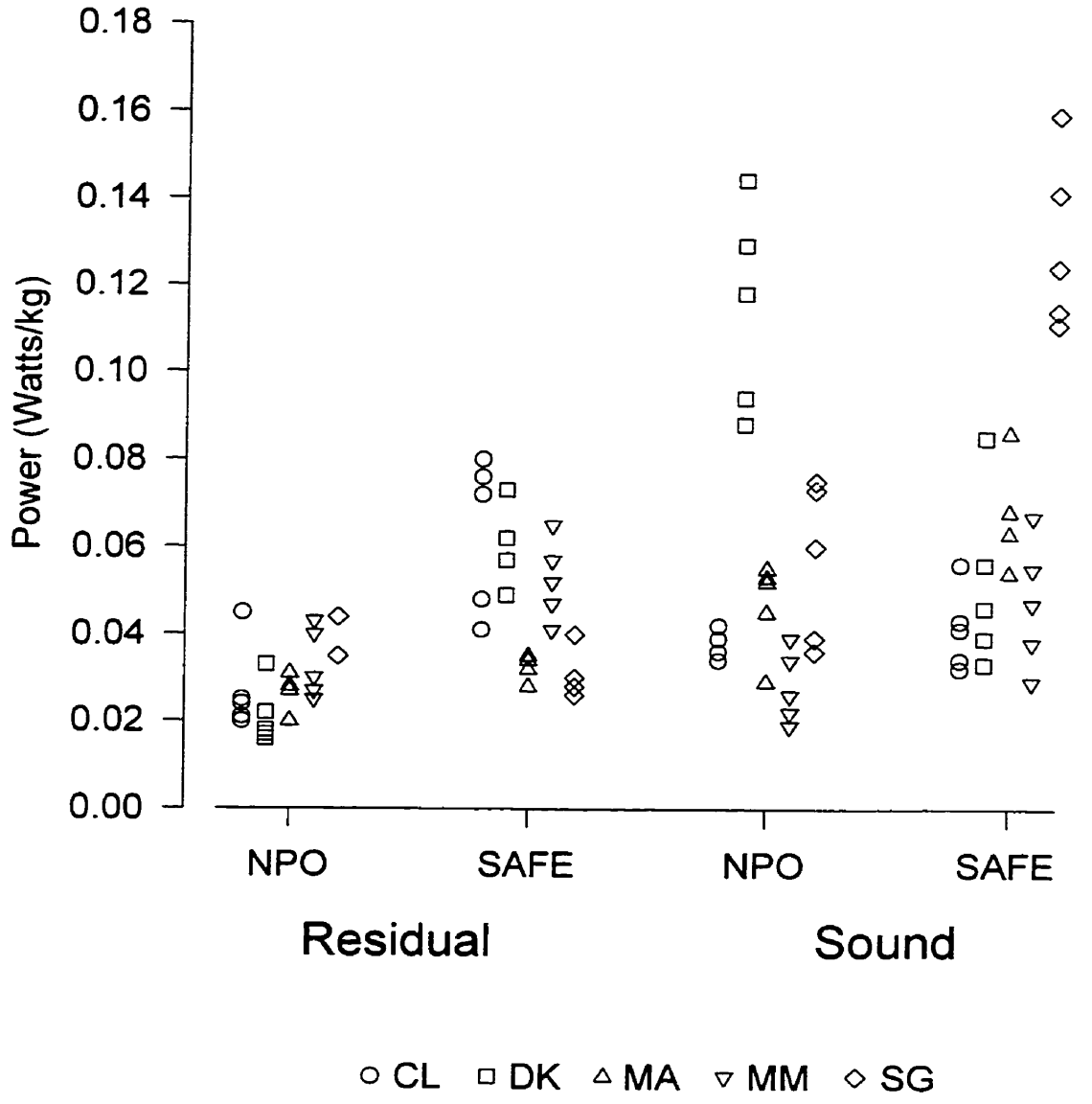


Figure H21: Maximum Power DP Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

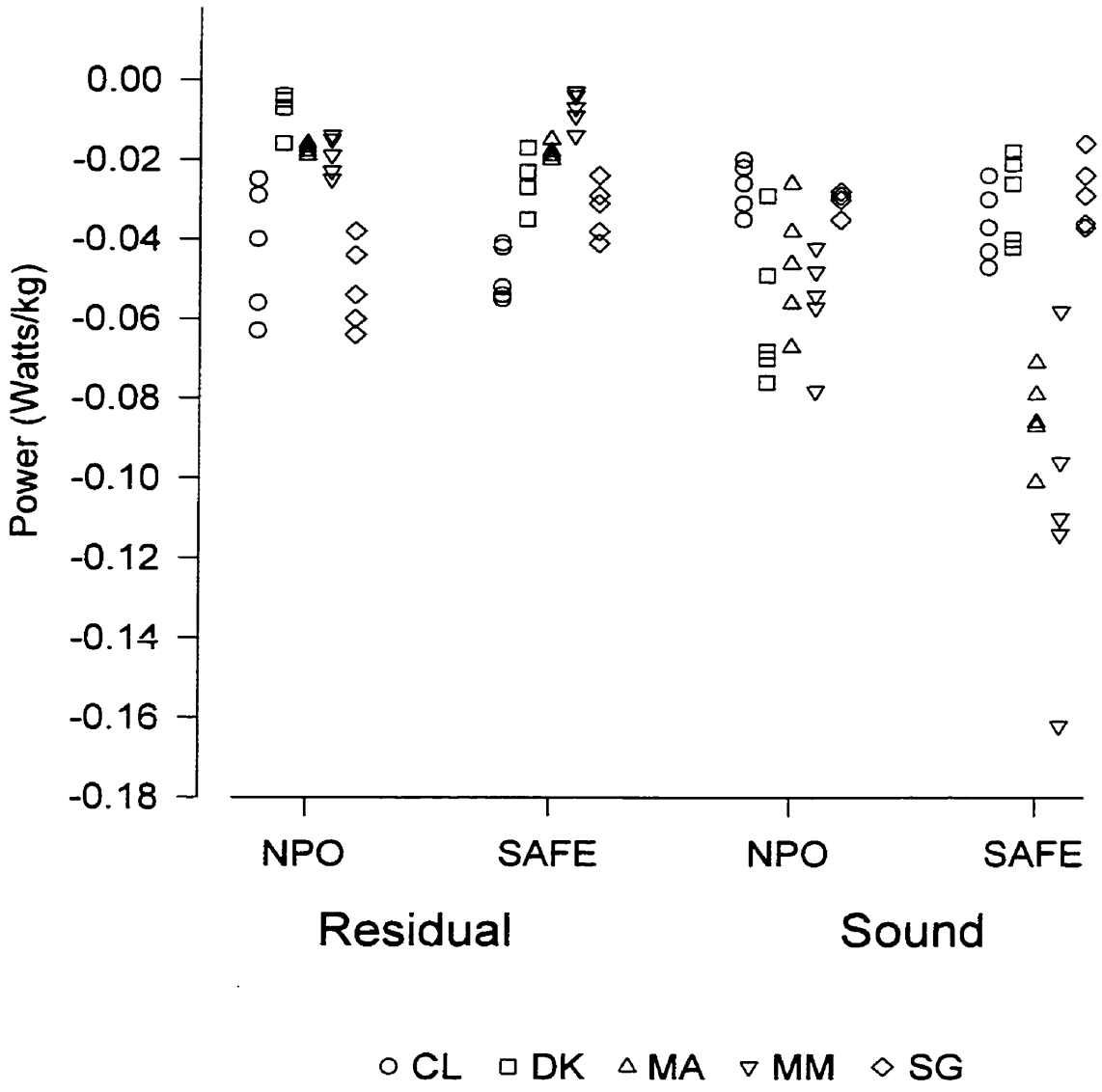


Figure H22: Minimum Power DP Axis (0-100% of Gait Cycle)
Knee Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

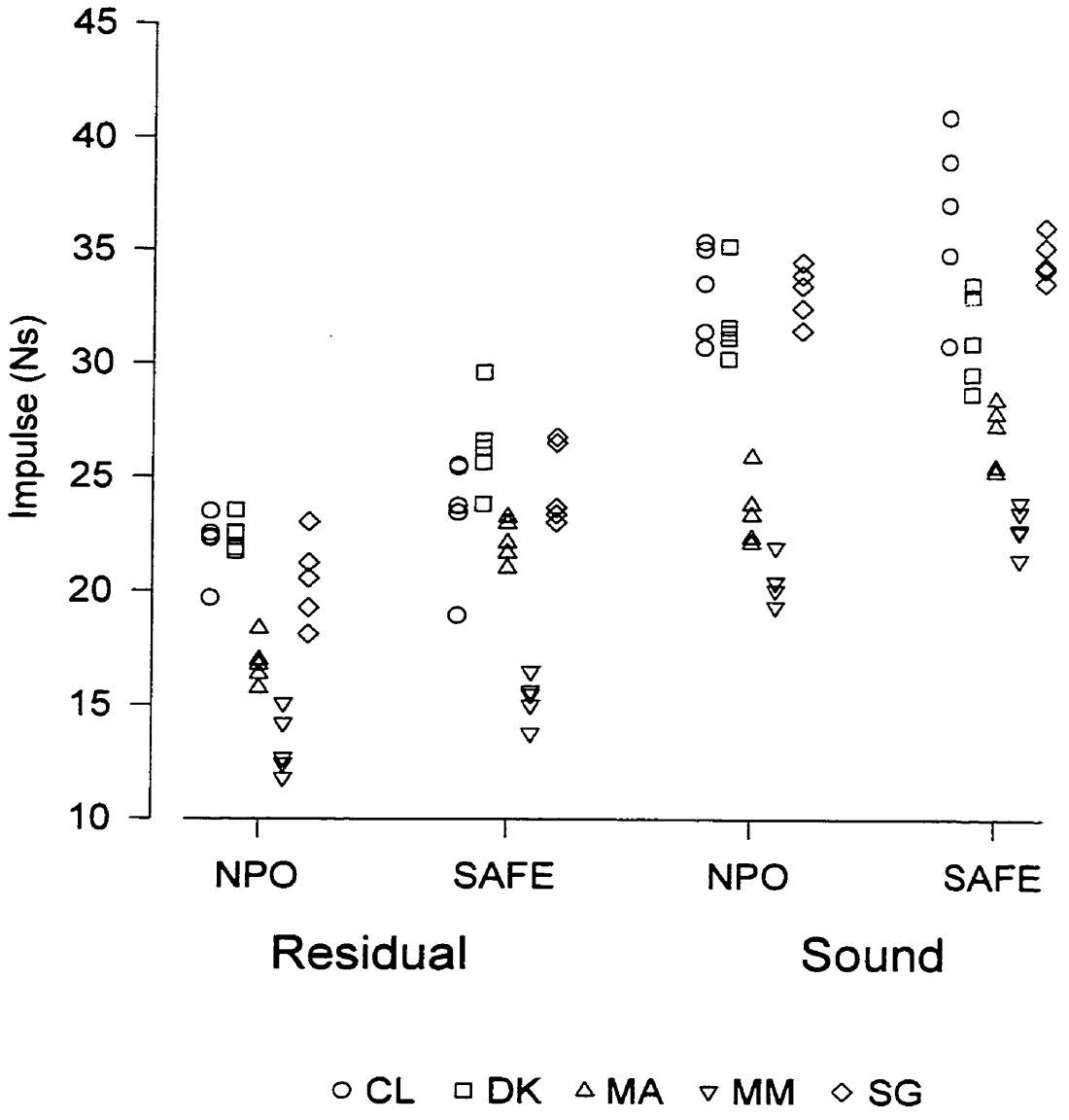


Figure H23: Positive Force Impulse PA Axis (0-100% of Gait Cycle)
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

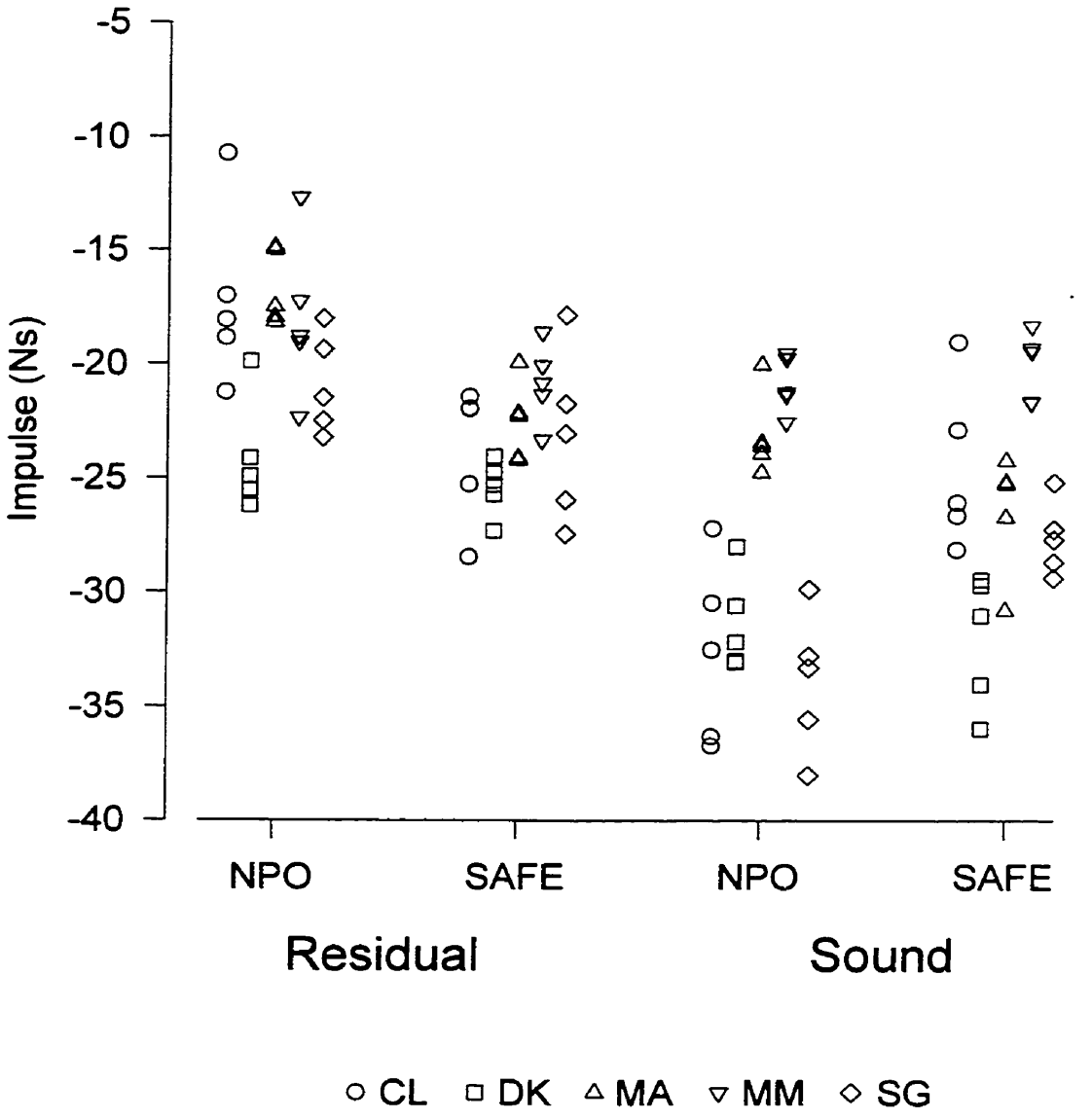


Figure H24: Negative Force Impulse PA Axis (0-100% of Gait Cycle)
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

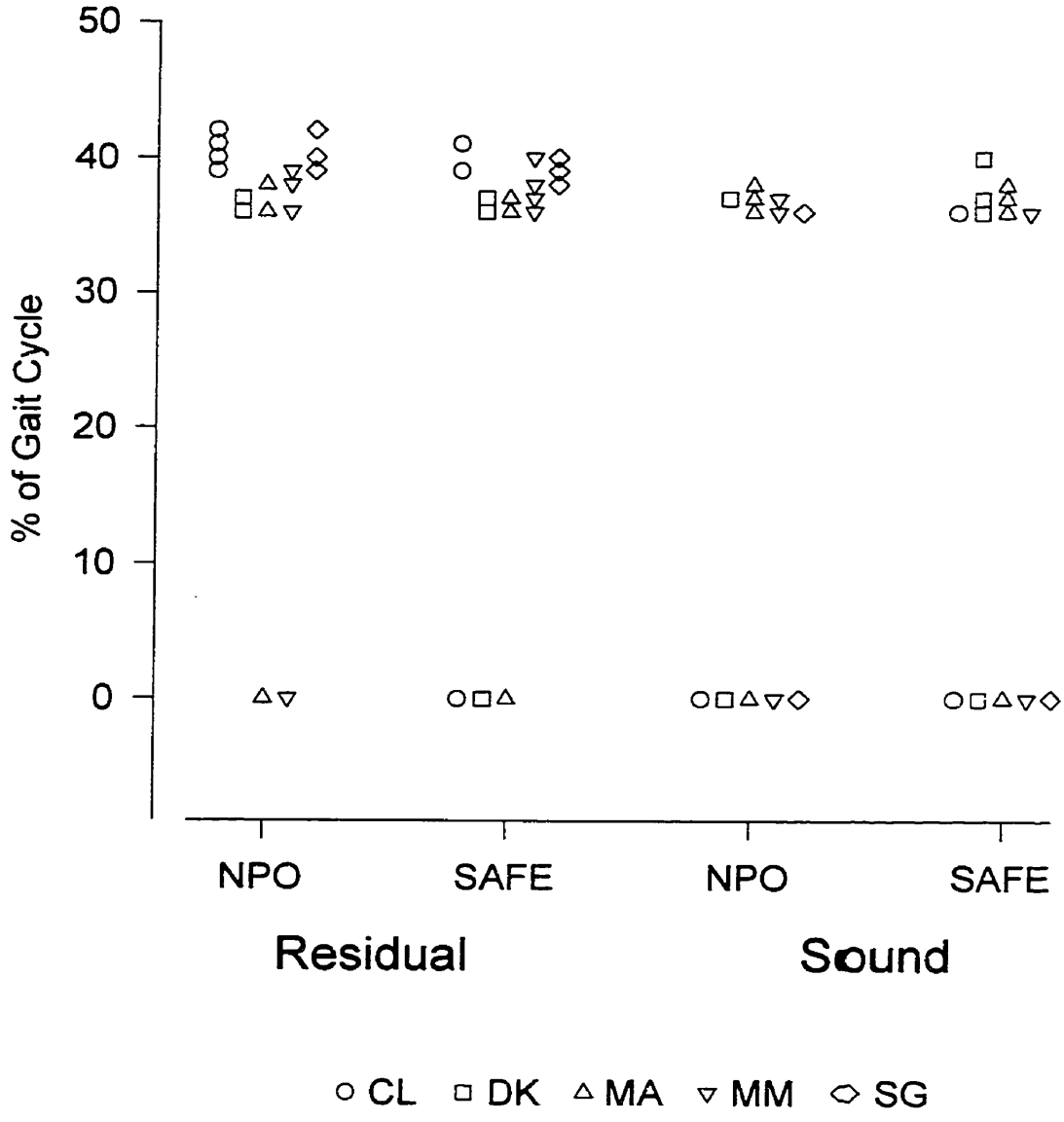


Figure H25: Location of Force Impulse Transfer PA Axis (20-50% of Gait Cycle)
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

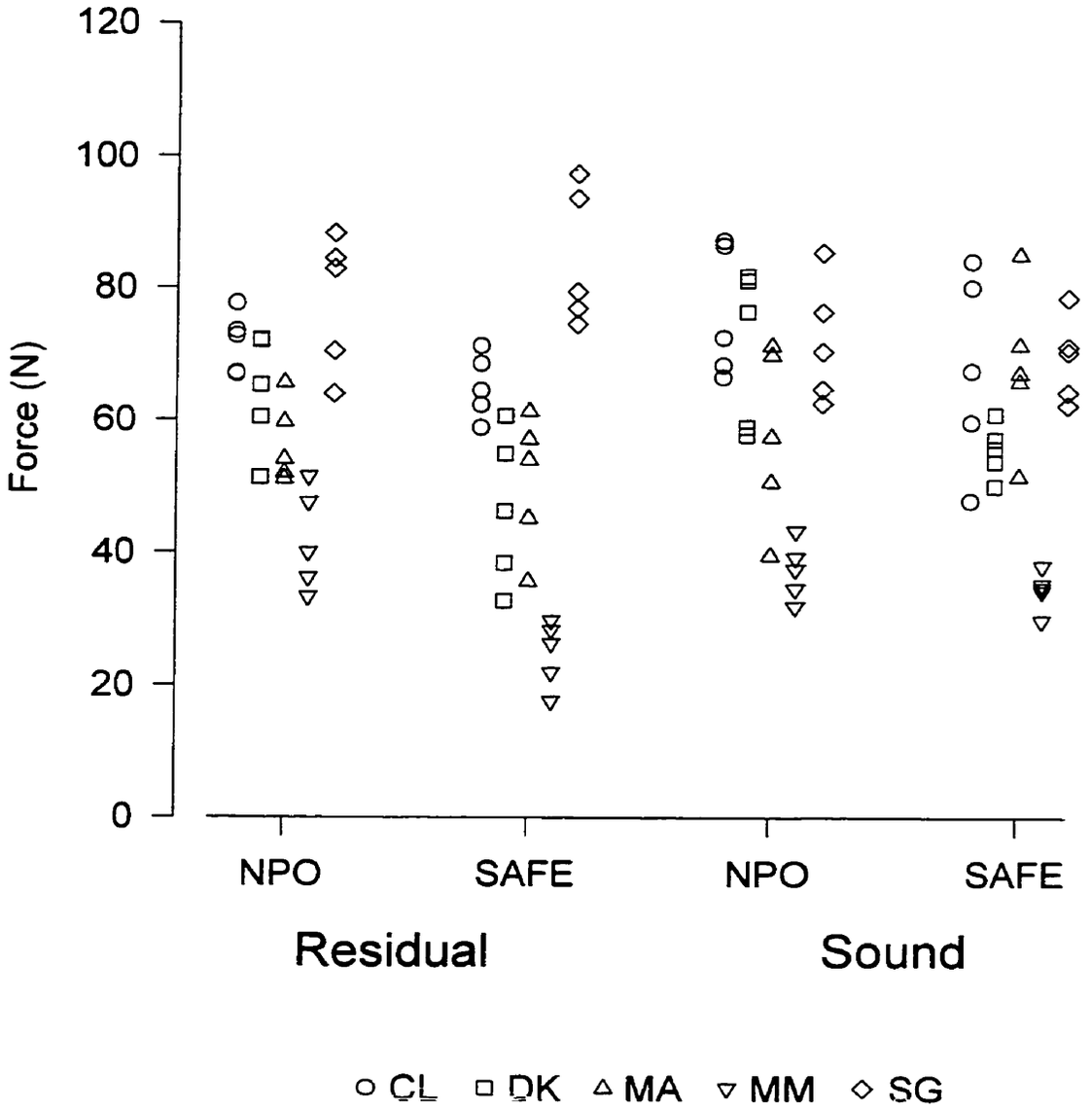


Figure H26: Maximum Ground Reaction Force LM Axis (0-30% of Gait Cycle)
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

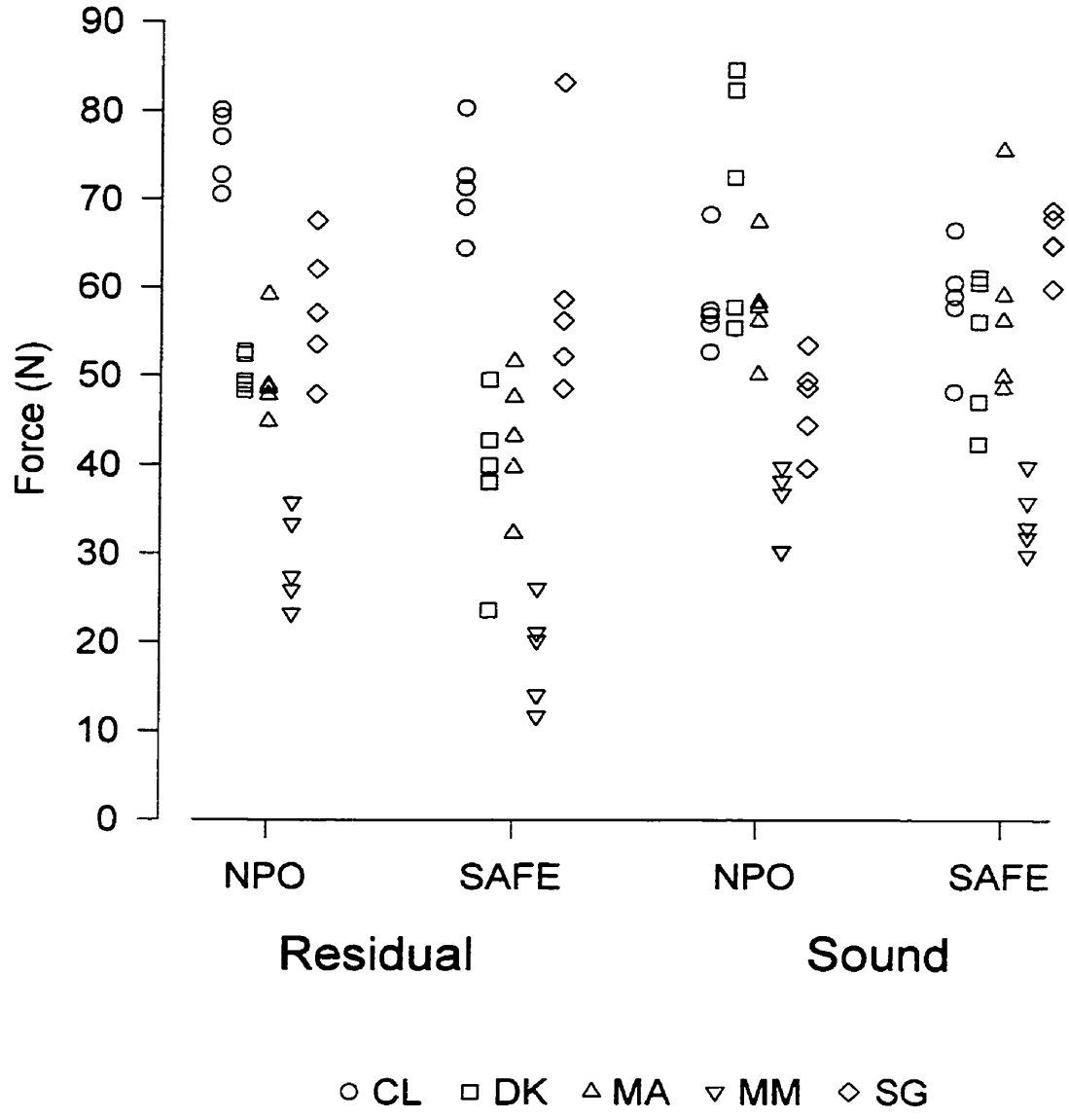


Figure H27: Maximum Ground Reaction Force LM Axis (30-60% of Gait Cycle)
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

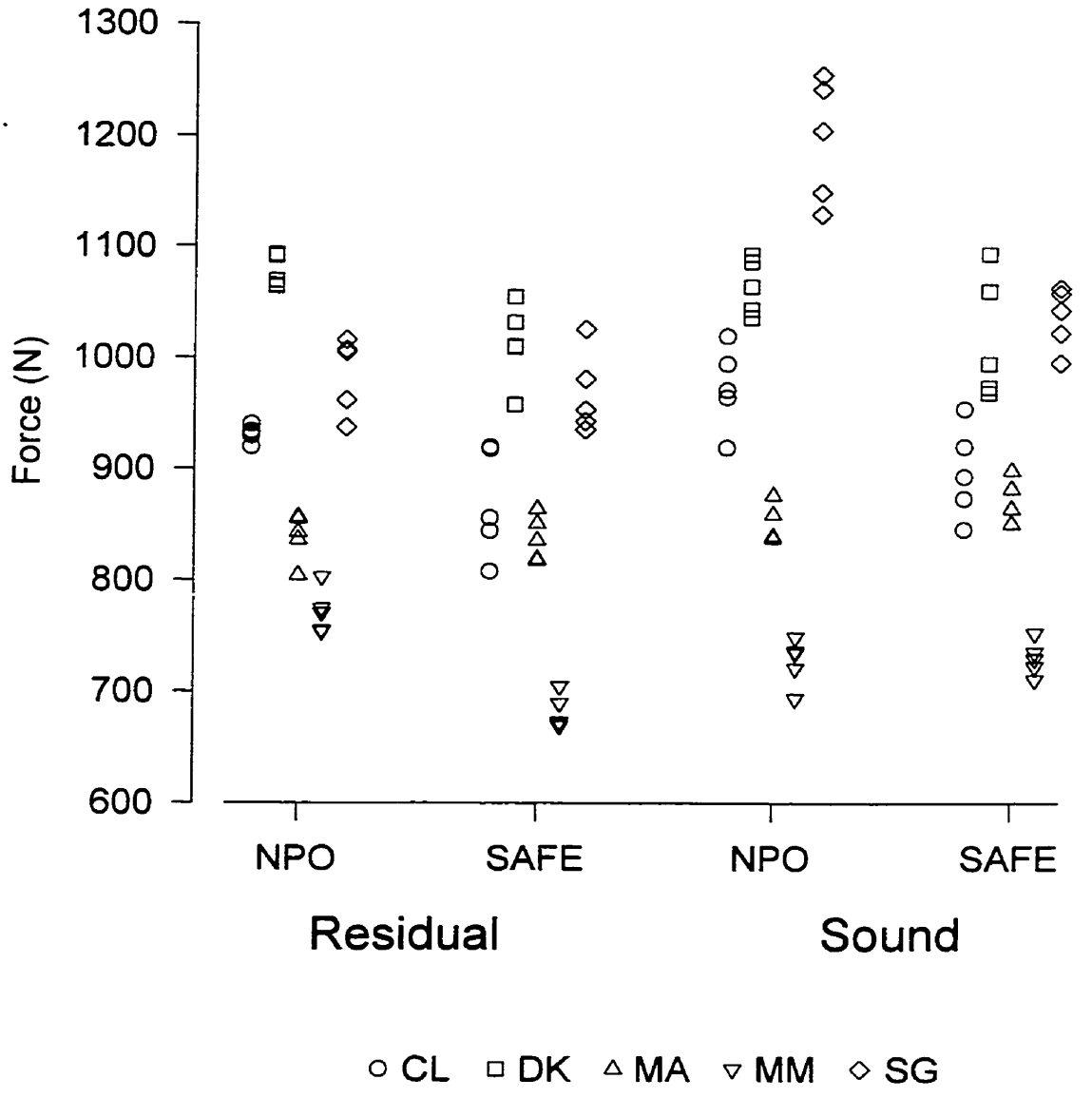


Figure H28: Maximum Ground Reaction Force DP Axis (0-30% of Gait Cycle)
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

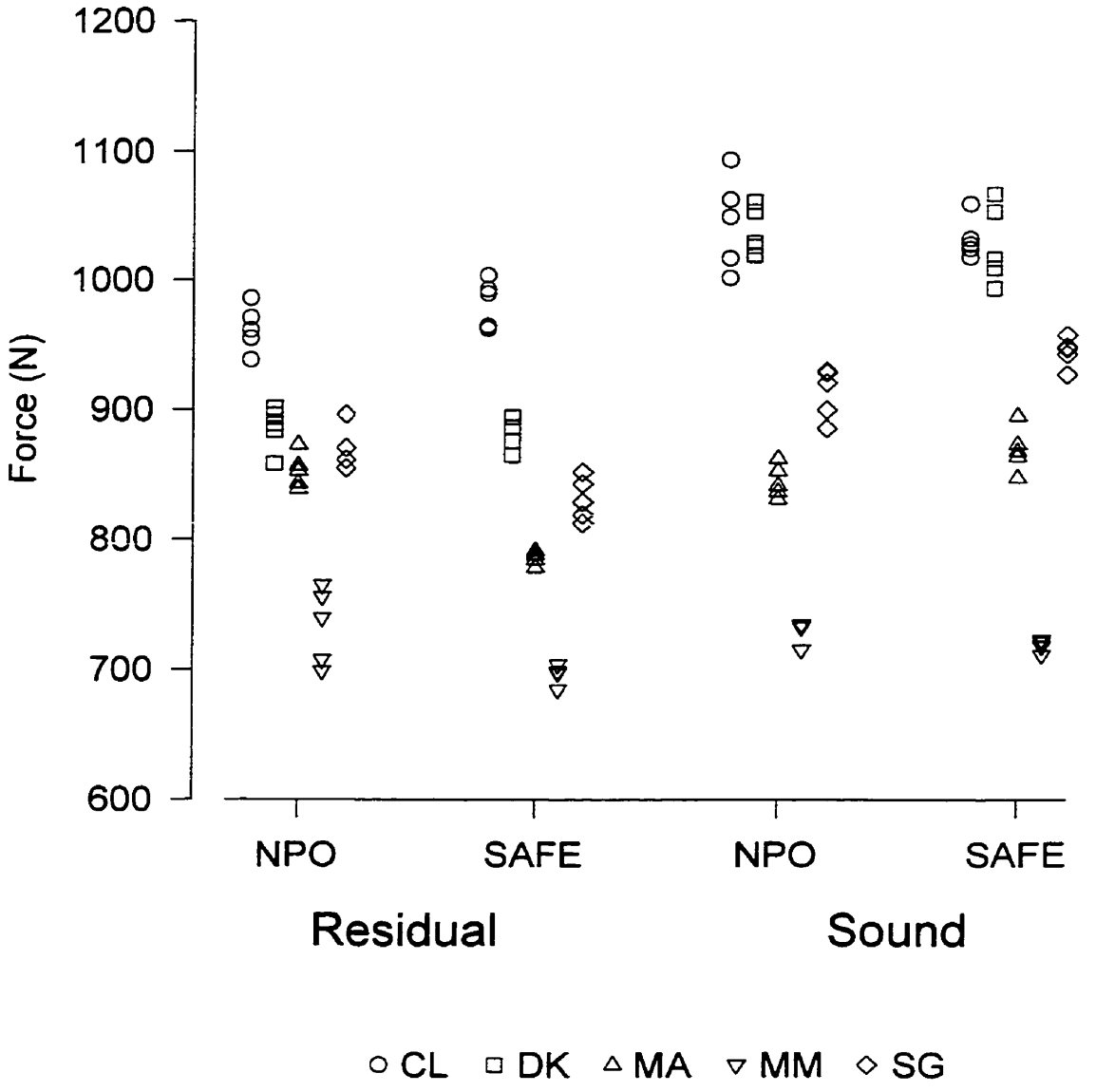


Figure H29: Maximum Ground Reaction Force DP Axis (30-60% of Gait Cycle)
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

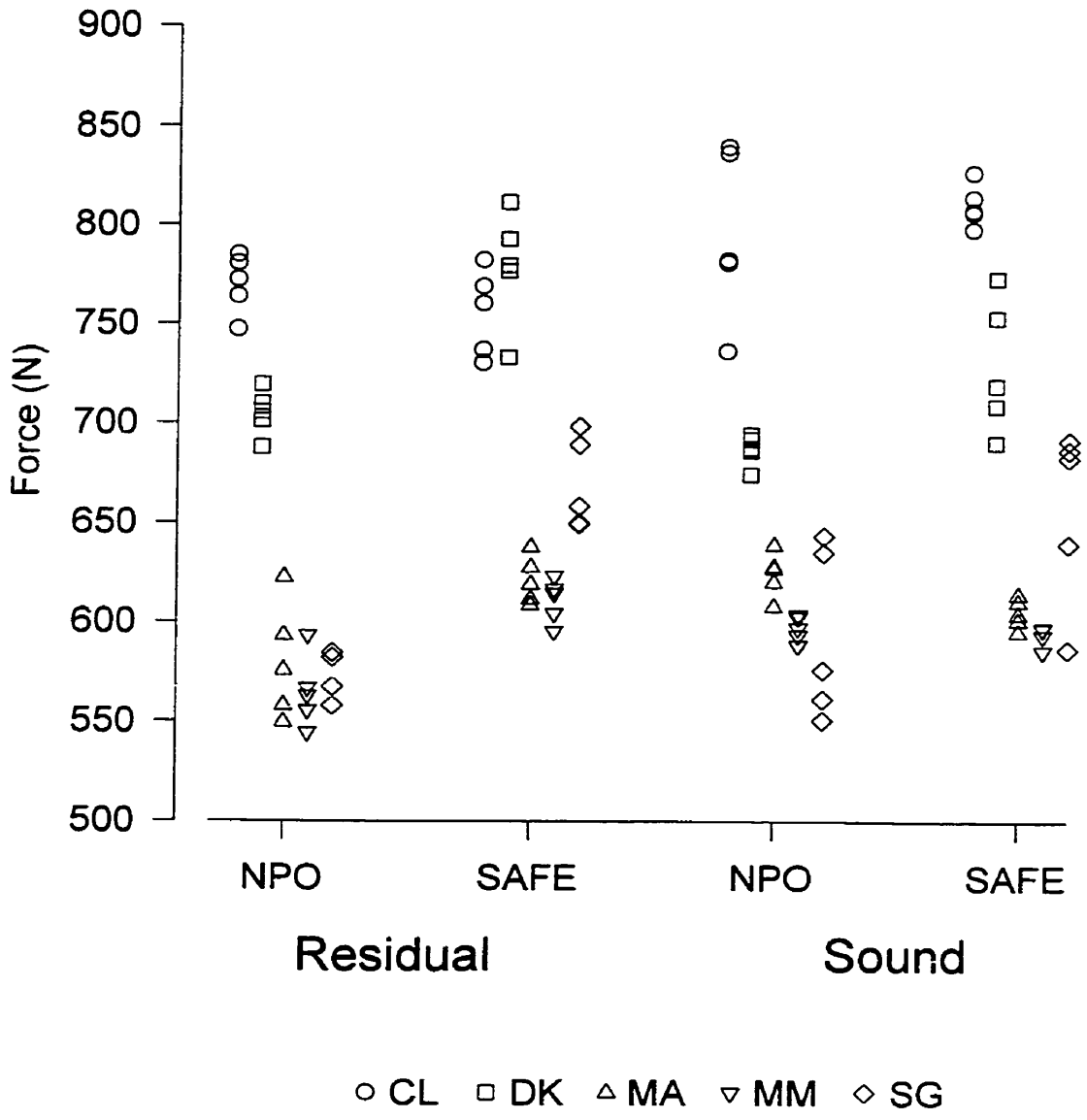


Figure H30: Minimum Ground Reaction Force DP Axis (20-50% of Gait Cycle)
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

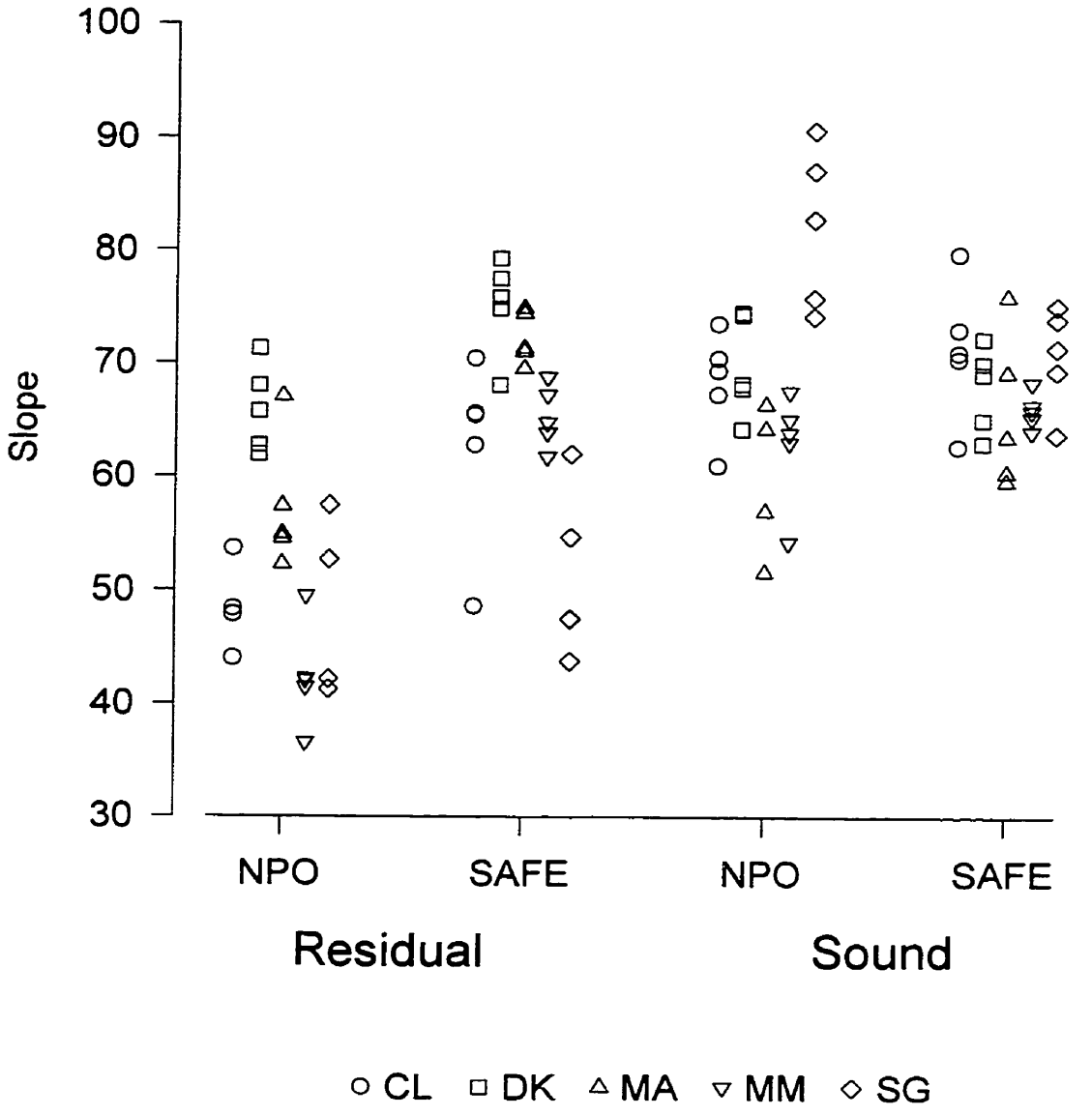


Figure H31: Slope of Ground Reaction Force DP Axis (0-11% of Gait Cycle)
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

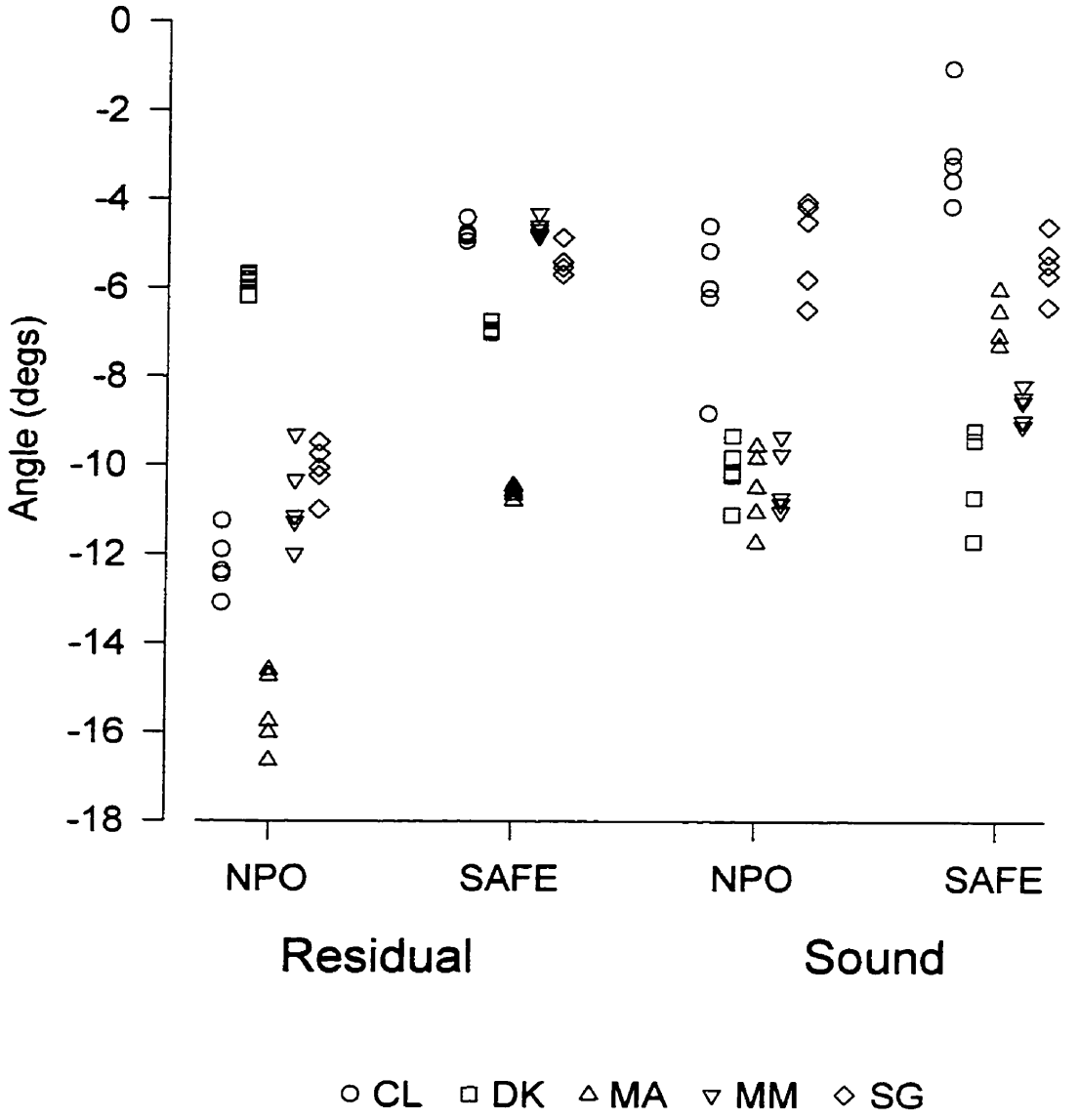


Figure H32: Minimum Angle DP Axis (0-20% of Gait Cycle)
Ankle Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

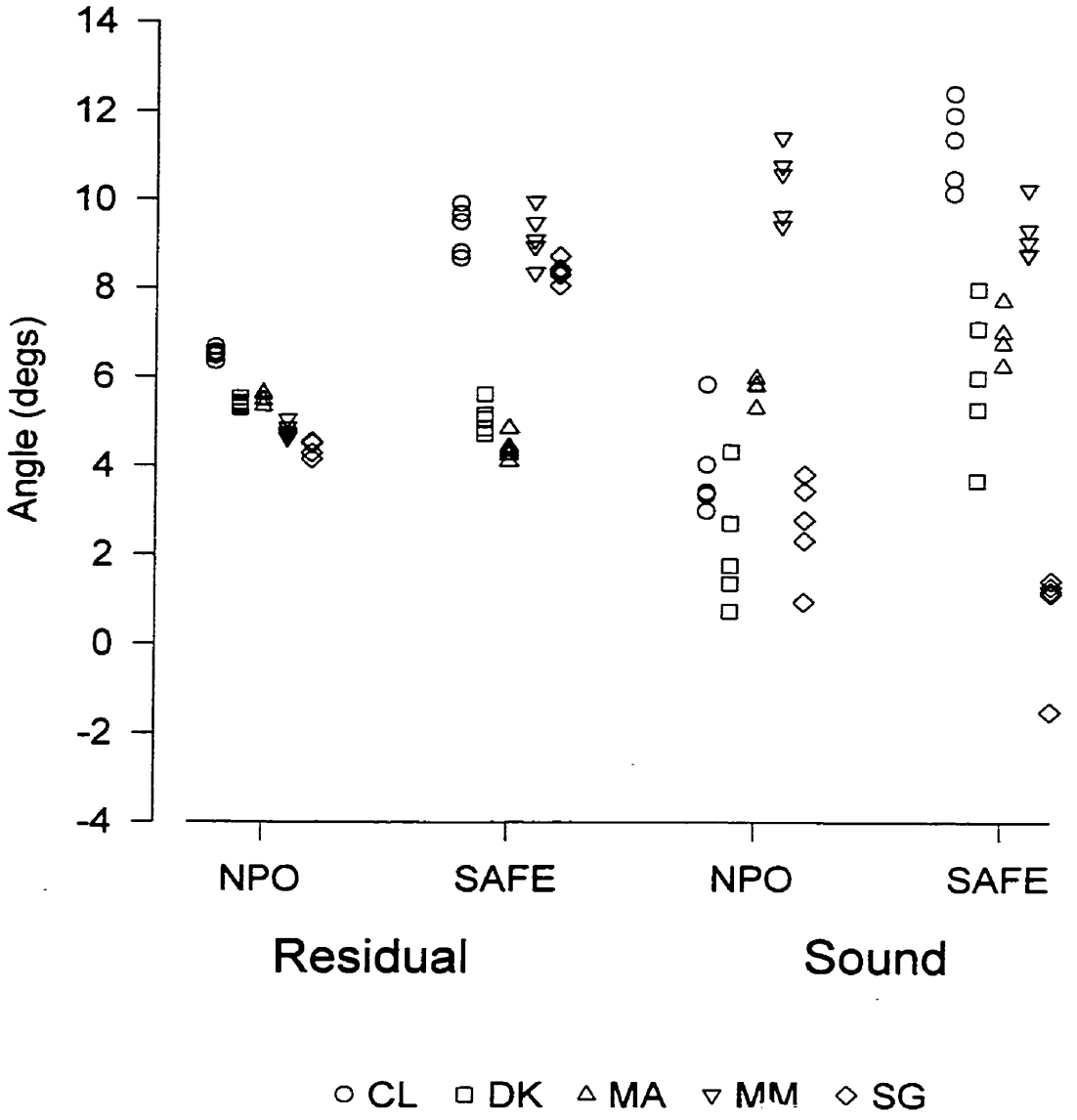


Figure H33: Maximum Angle DP Axis (10-60% of Gait Cycle)
Ankle Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)

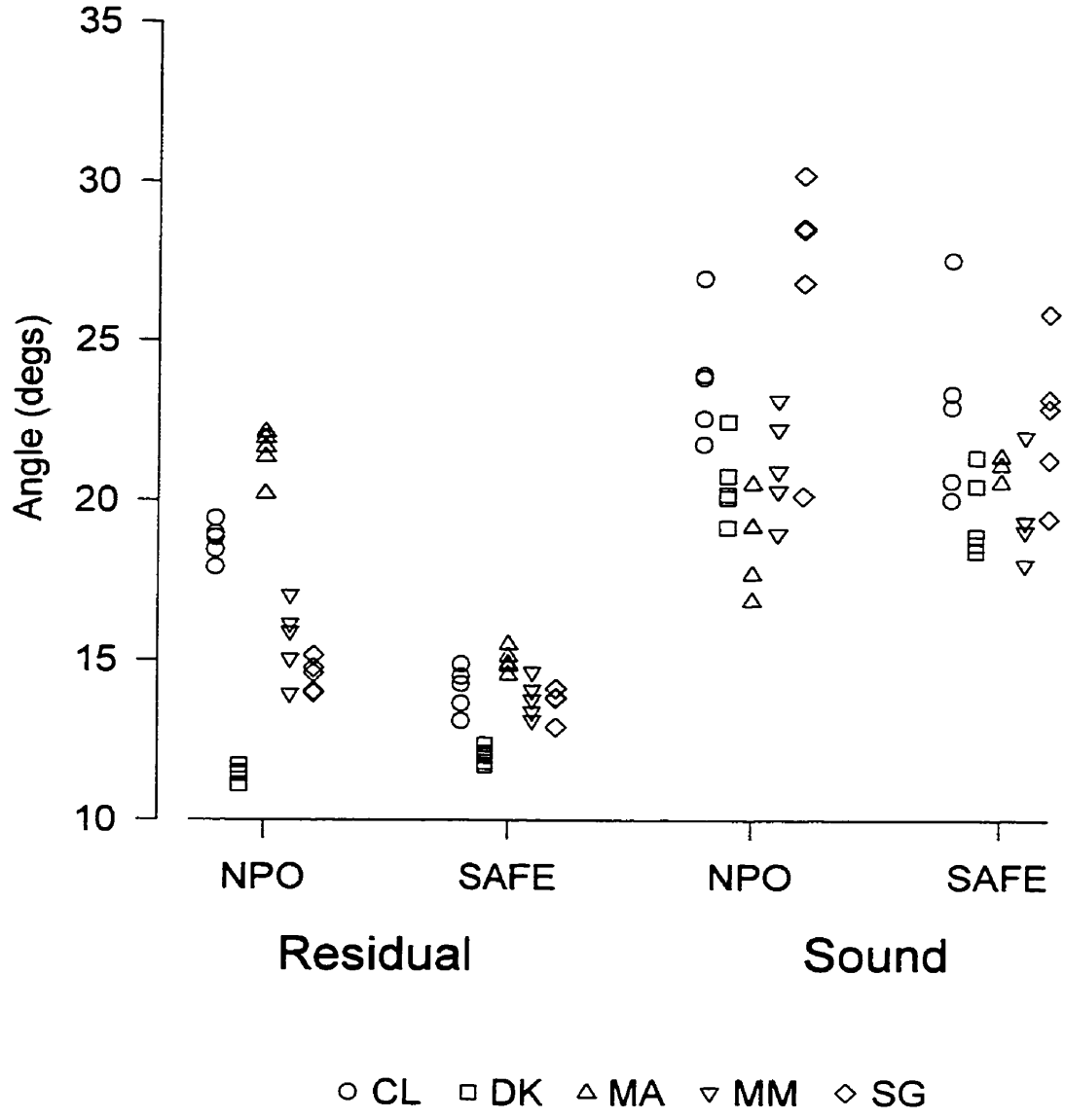


Figure H34: Range of Angle DP Axis (0-100% of Gait Cycle)
Ankle Joint
Five trials per subject
By limb (Residual, Sound), by prosthetic (NPO, SAFE)