THE ENERGETICS OF TRANSTIBIAL AND TRANSFEMORAL AMPUTEES WALKING ON TITANIUM AND STAINLESS STEEL PROSTHESES.

By

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This work is dedicated to a friend who never had the chance
to fulfill her dreams and to my family, whose
accomplishments are always an inspiration
for me to do my best.

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ABSTRACT

Several studies have been devoted to the metabolic costs of amputees walking on prostheses with different masses added to their components. However, limited study has been directed at quantifying the mass differences of the actual materials available to amputees and the metabolic and mechanical work required to walk on these materials. The energetics of two materials currently used in the design of lower extremity prosthetics were examined in an attempt to determine if mass differences had an effect on amputee walking. A total of fifteen, unilateral amputees (8 transfemoral and 7 transtibial) performed treadmill walking on prostheses assembled from titanium and stainless steel components. Standardized components (knees, pylons, adapters, feet) made from each material were added below the level of the socket. Submaximal oxygen consumption (W/kg) and mechanical power allowing transfers within and between segments (W/kg) were calculated as subjects walked at self-selected velocities until steady state was achieved. Results show that despite significant mechanical differences [$F(1,12) = 4.85$, $p<.048$], the decreased mass associated with the use of titanium materials does not have an effect on the metabolic costs [$F(1,14)=1.45$, $p<.249$] of the subjects in

this study. In addition, stride rate and stride length showed little differences when walking with both materials. Further division of subjects by age and experience walking on a prosthesis do suggest that older amputees and established walkers do benefit most from the use of titanium, both metabolically and mechanically. The choice of materials for use in every day walking will display differences in the mechanical work of amputees however, these differences are not great enough to realize metabolic consequences.

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CHAPTER I - INTRODUCTION

The question as to what may be the most efficient combination of mass and materials to go into the design of lower extremity prostheses has yet to be answered. Recently, there has been an increased use of strong yet, lightweight materials such as titanium and graphite in structures involving efficient motion. Consequently, there has been an increase in performance when these structures are powered by humans. The benefits are clearly evident in high performance equipment ranging from bicycles and tennis racquets to golf clubs. Similarly, rapid progress has been made to implement the use of titanium for prosthetic componentry.

With the addition of titanium, manufacturer's boast the benefits of decreased mass as it relates to efficiency and most patients testify in support of these statements. However, there is little scientific evidence to either justify or refute the value and increased pricing associated with lightweight materials. Although it would seem logical to expect increased efficiency with lower mass materials, it would also seem prudent to quantify this with experimental data.

The greatest benefactors of research into prosthetic devices are of course, the amputee's themselves however, investigation into the effectiveness of lightweight prosthetic componentry on the energetics of locomotion can assist both industry and clinical personnel. The benefits to industry would be to identify those components of prosthetic design that are not significantly related to energy savings (unnecessarily expensive to produce) and to better advise clients of expected benefits to be enjoyed by those components that are significantly related to energy savings or other performance measures. Clinical prosthetists and physiotherapists would value the results of such a study to aid in more accurate prescriptions of components and rehabilitative measures to better match the individual needs of their patients.

In addition, other professionals that would gain from this type of study include government officials and surgeons. Decisions by ministries of health regarding the appropriate amount of reimbursement to patients and possibly, for certification standards, could be improved by the information discovered. Orthopedic surgeons may use the findings to assist in decisions regarding appropriate levels of amputation.

The various studies used to determine the mechanical and metabolic properties of amputee gait while wearing differently massed prostheses have failed to provide any direct information on the current materials available to amputees (Bach et al., 1994; Czerniecki et al., 1994; Gailey et al., 1994; Hale, 1990; Skinner and Mote, 1989; Tashman et al., 1985; Menkveld et al., 1981). The mass differences and their contributions to energy costs accompanying the use of these materials for components remains unknown. The data that does exist can be very useful for future fabrication of materials and design of prosthetic devices however, it would seem logical to better understand the energetic properties of the current materials as a starting point. A study including the evaluation of these materials beyond that of strength and durability would certainly be useful to the manufacturers and distributors of such devices.

An evaluation of the energy savings, would depend on the biomechanical and physiological responses to locomotion by the people that commonly use the materials for walking. For a majority of individuals in the general amputee population (transtibial, transfemoral, young and old), the materials of choice and more importantly, the materials available to them are stainless steel and titanium. In the literature there has yet to be a comprehensive investigation into the inertial characteristics of prostheses assembled from stainless steel or titanium and the energetic effects of these characteristics on amputee walking.

The purposes of this study then, were as follows:

(i) to determine the inertial parameters of transfemoral (TF) and transtibial (TT) prostheses assembled from stainless steel and titanium components;

(ii) to quantify the the total mechanical power and metabolic costs of TF and TT amputees at various ages while walking on prostheses assembled from stainless steel and titanium components.

The hypotheses of this study were:

(i) that the prostheses worn by TF and TT amputees will have greater masses when assembled from stainless steel components in comparison to prostheses assembled from titanium components;

(ii) that the moment of inertia of the prosthetic limb about the center of rotation of the knee will be greater for prostheses assembled from stainless steel components in comparison to prostheses assembled from titanium components;

(iii) that the energy costs and mechanical work of locomotion in lower extremity amputees will be greater in subjects that walk using prostheses assembled from heavier components;

(iv) that the energy costs and mechanical work of subjects will be greater at higher levels of amputation due to increased componentry mass;

(v) that the effects of age and walking experience on a prosthesis will reduce the energy costs and mechanical work of older, established subjects using titanium prostheses as compared to stainless steel.

CHAPTER II - REVIEW OF LITERATURE

In order to comprehensively quantify and evaluate the energy expenditure of amputees walking wearing different massed prostheses several contributing factors must be considered. These include, measurement of oxygen consumption, amputee economy and, the effects of segment mass alterations during normal walking. In addition, knowledge on the methods suggested for calculating mechanical work is essential. Lastly, investigations involving alterations in prosthesis mass may provide the most beneficial information. These topics will be reviewed as well as related research on the kinematics, kinetics, mechanical and metabolic work of amputees walking wearing different massed prostheses and the determination of amputee and prostheses inertial characteristics.

Oxygen Consumption

One method of evaluating the metabolic energy expenditure of normal gait or of an amputee walking on a prosthesis is to determine the oxygen consumption of the individual. Furusawa (1924) helped to establish this fact stating that, "a careful investigation of the oxygen requirement of different types of effort and of different speeds and manners of working may prove valuable in a scientific study of human muscular activity". Furthermore, Williams (1985), commented that oxygen consumption has generally been accepted as a convenient and accurate method of assessing metabolic energy costs as long as the workload remains aerobic. Thus, during steady rate exercise, the oxygen consumed closely mirrors the energy expended. The oxygen consumption of performing a task at a submaximal level has been defined as economy and it is generally agreed that the lower a submaximal oxygen uptake for a given workload, the greater the economy (Cavanagh and Kram, 1985a). This notion of economy is the preferred term to be used as an indicator of physiological performance.

Amputee Economy

The ability to walk wearing a prosthesis following the amputation of a lower limb becomes a very difficult and complicated task. With time, the amputee develops the psychological and physical strength as well as the motor abilities associated with rehabilitating the residual nervous supply and muscles to achieve control of a prosthesis. This usually results in a gait pattern that ensures the amputee safety and stability but, does not always reflect economical locomotion.

As the amputee becomes more confident he/she becomes more active, performing more physically demanding tasks and therefore, strives for improved economy of movement. These desires are usually accommodated by a new prosthesis with differing components. One way to monitor the progression of an amputee or to evaluate their performance on a newly designed prosthesis is to measure oxygen consumption. This information can provide insight into whether or not the amputee is walking with greater economy.

It is believed by Waters et al. (1992) that, metabolic measures provide global information on overall gait performance and a means of quantifying the overall physiologic penalty resulting from pathological gait. Waters et al. (1976) used this thinking in collecting steady state walking data of 70 traumatic and dyvascular amputees. Amputees at the Syme's, transtibial and transfemoral levels were tested and the data indicated that performance was significantly better the lower the level of amputation. In a study of 37 unilateral, above-knee (AK) amputees by James (1973) it was determined that the prosthetic gait of the amputees was found to be metabolically more costly than the gait of a reference group of 26 "normal" walkers. In both studies, it was found that there was a maximum aerobic capacity that was significantly lower in AK's as compared to below knee's (BK's) and normal subjects. It is speculated that the data on maximum aerobic capacity is a reflection of the AK's adapting their lifestyle to a less active one that results in reduced physical conditioning of the muscles and

In addition to the two studies mentioned above, several other small investigations with limited subjects involving economy measurements on amputees at various levels have been summarized by Fisher and Gullicson (1978) resulting in some general conclusions. It was concluded that the average BK amputee walks 36% slower expending 2% more kcal/min and 41% more kcal/m than the normal person and, the average AK amputee walks 43% slower and expends 5% less kcal/min and 89% more kcal/m than the normal person.

Lastly, with regards to oxygen consumption some studies do exist that attempt to determine the effectiveness of certain prosthetic components by measuring the physiological responses during amputee walking. Such studies include economy comparisons of quadrilateral and CAT-CAM sockets (Gailey et al., 1993) and prostheses with open and locked knee mechanisms (Isakov et al., 1985). Thus, the measurement of oxygen consumption during amputee gait proves to be an often used tool for explaining differences in economy amongst the amputees themselves (at different levels), against normals and, as an evaluator of components. Each of these assessments of amputee economy is important to this study as they provide information that can be anticipated when analyzing group outcomes and in the latter case, to substantiate the use of such tests for learning more about the prosthetic devices themselves.

Normals "Loading" and Economy

As information regarding added mass effects to prosthetic limbs is limited, insight into "loading" studies of normals may be useful in addressing amputee economy. A number of studies have been devoted to both mechanical and physiological responses to extremity loading during normals running in an attempt to quantify the value of using hand held or ankle weights as conditioning assets (Cavanagh and Kram, 1989; Cavanagh and Kram, 1985b; Claremont and Hall, 1988; Martin, 1985; Myers and Steudel, 1985; Bhambhani et al., 1989). Fewer studies have examined normal walking and added mass effects on economy however, at least in one case the aim was to collect some data that may lead to more accurate prescriptions of prosthetic and orthotic devices (Skinner and Barrack, 1990). Skinner and Barrack found an increase in oxygen consumption (in ml/kg/min) of 6.3% and 14.2% respectively, for normals walking at steady state with 1.82 kilograms (kg) added asymmetrically (to one ankle) and symmetrically (to both ankles). Only the symmetrical loading conditions were significant. Also, cadence and stride length remained unchanged using the same weight addition protocol. Thus, by applying normal values to amputees it appears that the asymmetrical loading that would be displayed by a unilateral amputee requires a greater addition of mass than 1.82 kg to affect oxygen consumption.

Perhaps this is true in BK's however, this mass value is misleading as it is an addition applied to a normal limb that usually already weighs substantially more than most prescribed prosthetic AK legs. Therefore, addition of mass to most prosthetic limbs would have to be much greater than the values described here in order to reach the same experimental weights as normals. Even with the appropriate additions, the lack of musculature in the amputee may provide different physiological results that may be more or less sensitive to changing mass than seen in normals. These results and others similar to those found in this study are valuable but should be interpreted with some caution when extending the findings from normals to amputees.

Other studies of this nature all include symmetrical loading to the feet of normals and do provide some indication of what may happen when loading the segments of amputees. For example, Miller and Stamford (1987) added masses of 2.25 kg's to each ankle of normals and observed a 0.8% increase in VO_2\100 grams added which agrees with most other walking and running studies that have averaged a 0.7% to 1.0% increase (Jones et al., 1984; Martin, 1985). These results also agreed with the earlier work of Ralston and Lukin (1969), and Soule and Goldman (1969) who observed larger increments in energy expenditure when loads are applied at the ankles in comparison to unloaded conditions.

In the case of Ralston and Lukin additional loads of 2 kg's on each foot translated to an increased metabolic expenditure by 31% over control groups.

The above information describes the metabolic behaviour of normals and for the purposes of this study should only be interpreted while considering the obvious differences in normal and amputee segment characteristics. However, these data are essential in understanding the relationship between uncommon segment masses and metabolic responses to them. After all, if one wants to provide a prosthesis that can accomplish normal function it should at least initially, be studied in the same manner as the target goal.

Mechanical Work

According to Winter (1990), the energy flows that cause movement are of paramount importance to any biomechanical study that addresses movement. He states, "valid mechanical work calculations are essential to any efficiency assessments that are made in sports or work-related tasks" (Winter, 1990). The meaningfulness of these statements are equally as powerful when applied to pathological movement that requires the use of artificial aids.

Calculation of Mechanical Work

Three methods commonly used to estimate mechanical work include: 1) work output based on kinematics of the center of mass (COM), 2) work based on kinematics of the individual body segments or 3) a kinetic approach based on segment dynamics and associated joint moments and forces (Robertson and Winter, 1980).

(i) Whole Body Center of Mass Models

The most basic method of estimating mechanical work involves changes in energy of the whole body COM. These methods only consider external work and ignore the work done by the moving limbs (internal work). Although, this is a relatively easy measure to take, it underestimates the total energy changes involved in the movement being studied (Winter, 1979). Total body mechanical work can be calculated, assuming complete exchange between potential and kinetic energy components of the COM or by assuming no exchanges if the absolute values of each component are used.

(ii) Segmental Models

Segmental analyses provide a measure of both external and internal mechanical work by identifying the changes in energy of each individual body segment. However, the number of ways in which the energy changes from all the segments and their summation for an overall measure or work presents some complications (Williams, 1985). Common to all methods, is the assumption that the instantaneous energy of a given segment is derived from potential and kinetic (translational and rotational) energies.

Differences exist when summing the changes in instantaneous energy. A method termed `pseudowork' was developed by Norman and co-workers (1976) which added the absolute changes in energy of each segment across segments and then over the whole locomotion cycle. Criticism of this method by others (Pierrynowski et al. 1980; Williams and Cavanagh, 1983; Winter, 1979) showed that it is likely to overestimate the actual mechanical work as there is no provision for transfer of energy. As a result, alternative methods accounting for transfer of energy within and between segments have been developed (Pierrynowski et al., 1980; Williams and Cavanagh, 1983; Winter, 1979). A method proposed by Winter (1979) attempts to account for energy transfers within and between segments, such that energy is transferred between segments when the total instantaneous energy of one segment decreases. Any energy lost is transferred through the system if another segment is increasing its total segment energy. This method has also received criticism as transfers which do not make "anatomical" sense may be allowed. This is the case when a decrease in total energy of the right foot occurs during the same time interval as an increase in energy of the left forearm. The equations in this model would assume that complete transfer of energy had occurred between two widely separated segments (Williams and Cavanagh, 1983). In addition, the metabolic cost of negative and positive work is not accounted for. An attempt to make adjustments for the relative metabolic cost of negative and positive muscular work was then included (Pierrynowski et al., 1980). Zarrugh (1981) assumed that the metabolic cost of negative work was zero in assessing the power requirements of walking, thereby reporting a lower boundary for mechanical power generated during that activity. It could also be suggested that an adjustment is needed to restrict between segment transfers to adjacent segments only. Use of energy transfer concepts in studying economy differences between individuals suggests an energy conservation mechanism which can provide valuable information however, some researchers (Martin et al., 1993) caution its use, questioning the physiological validity of the energy transfers and recommending further study in the area.

(iii) Kinetic Methods

In an attempt to reduce the error that may be produced in kinematic models Robertson and Winter (1980), proposed a kinetic approach. Net joint forces and moments were used to compute net joint and muscle moment powers. The joint power can be considered to be the rate of mechanical energy transfer between adjacent segments. Under the assumption that, the use of net moments gives an indication of the mechanisms used by the muscles to generate and dissipate power, it was thought that the values calculated using this method would be more closely related to metabolic cost values, as compared to kinematic measures. The disadvantage of this method is that the joint and muscle power calculations require a force platform which is not possible in treadmill protocols required for steady state economy calculations.

(iv) Factors Not Accounted For In Biomechanical Models

The use of biomechanical models may lead to unrealistic estimates of mechanical work. Underestimations result as these models cannot account for co-contraction of antagonistic muscles, isometric work against gravity and non-sagittal plane motion. Factors that are not included in biomechanical models and thus, lead to overestimates of mechanical work include, passive tissue contributions and utilization of stored elastic energy. However, results of a recent study suggest that applying a biomechanical model using the kinematic approach allowing energy exchanges within and between segments of the same limb does demonstrate good relationships between mechanical work and metabolic cost rates within children, when running at different speeds (Dowling et al., 1995). This showed that despite not having a force platform and energetic factors not well accounted for, the present biomechanical models have predictive and therefore, diagnostic value. Thus, it is not unreasonable to expect that such a biomechanical model would be able to yield mechanical reasons for observed metabolic cost differences of amputees walking on prostheses assembled from different materials.

Alterations in Prostheses Mass

(i) Kinematic and Kinetic Studies

Several studies have examined the kinematics and certain kinetic variables of amputee gait with respect to varying prostheses mass however, few have quantified the mechanical work performed by amputees under such conditions. The kinematic and kinetic studies that exist focus mainly on AK amputees and indirectly address the issue of mass via computer modelling and by the addition of externally applied weight. Mena et al. (1981) and Menkveld et al. (1981) used a three segment model of the leg and found that the desired simulated leg motion was less sensitive to increases in segment inertial characteristics than decreases. Using predictions from the model it was observed that the addition of 5 pounds (lbs.) at 12" proximal to the ankle of an actual AK subject, decreased the swing leg thigh torque and increased the stride length and velocity. Tsai and Mansour (1986) developed a two segment AK model which included knee unit control and used specified normal hip trajectory and moment as input and found that lightweight prostheses generally gave greater deviation from normal kinematics than heavier models. However, it should be recognized that normal kinematics, trajectories and moments were used to drive the simulation. This makes the assumption that these control variables of amputee gait remain unchanged when compared with normals. It should also be noted that no combination of inputs minimized deviation from normal kinematics.

Manipulating the shank of one AK amputee to compare center of mass locations of 18.7 and 31.7 centimeters (cm) distal to the knee joint allowed Tashman et al., (1985) to observe a slightly higher cadence, stride length and walking speed for the proximal mass. This reduced gait asymmetry for this individual. Hale (1990), loaded the shank-foot components of six AK amputees up to 100% of the sound shank mass and found that free walking speed went unaffected however, the hip muscular effort of swing phase significantly increased as load increased mainly during deceleration and preparation for the upcoming foot strike. Despite this, four of the six subjects preferred a loading of 75% that of the sound shank.

The effects of added mass to the prosthetic foot have also been tested for kinematics and kinetic characteristics. Donn et al., (1989) examined the gait of ten BK amputees with added plasticine masses ranging from 50

- 200 grams, just anterior to the prosthetic ankle joint on a standardized shoe. Total foot masses ranged from 260 - 460 grams. Various parameters were measured leading to an estimate of swing phase symmetry which was correlated with the subject's "preferred" shoe mass. Results indicated that, at the preferred shoe mass (139 - 318 grams), symmetry was greatest (correlations 0.78 - 0.81) in comparison to a poor correlation when subject's wore the lightest shoe with no mass applied (121 - 325 grams). The authors concluded that BK gait symmetry suffers with lightweight footwear.

In a study with similar methods, Godfrey et al., (1977) taped 113.4 and 226.8 gram weights to the prosthetic foot to examine the walking of six AK amputees wearing six different types of knee joints. No differences were observed for the pace rate, average horizontal velocity, and heel rise velocity of subjects walking with no mass applied or under weighted conditions.

(ii) Mechanical Work

The limited number of studies that do exist addressing the mechanical work performed by the amputee while walking wearing different massed prostheses also employ modelling and added weight designs. Beck and Czerniecki (1994) constructed a hypothetical subject from averaged measurements on four normal subjects for the basis of constructing a model. This simulation was used to assess a broad range of inertial characteristics and their influence upon required knee power, muscle work at the hip and hip joint energy transfer during swing phase. Their data indicate that if an increase in energy flow to the trunk with minimal increase in hip mechanical work and positive joint power is desired, a heavier shank with a more proximal (to the knee) center of mass should be used. These authors do caution that this arrangement may not necessarily correlate with metabolic cost. Van De Veen et al., (1987) used a bondgraph model to observe a decrease in required power during AK swing phase when the total mass of the prosthesis is decreased. The greatest changes were observed when mass was eliminated from the prosthetic limb model at distal locations (shoe and foot).

Bach et al., (1994) measured the total mechanical power assuming energy exchange within and between segments, of five AK amputees with a combined simulation - added mass study. No significance was found in any of the treatment conditions during actual amputee gait including an optimal energy mass and optimal energy location condition as prescribed from an optimization simulation. In addition, the mechanical power calculated correlated very poorly with metabolic power measurements. In an earlier study with similar dependent variables Skinner and Mote, (1989) determined two measures of mechanical work, one based on power developed across the joints and one based on body segment kinetic and potential energy changes however, these values were only evaluated as predictors of metabolic energy expenditure and not separately. These variables were rejected as being significant correlates with metabolic data.

In summary, kinematic and kinetic assessments on the topic of changing prosthetic mass display some inconsistency in both designs and outcomes. Some constant findings include the use of computer simulations or models, the applications of external weight from model predictions or from clinical estimates and the examination of AK subjects only, with the exception of one BK study by Donn et al., (1989). Each study tested a rather small number of subjects ranging from one to ten, with most studies including four to six subjects and some including no actual amputee walking trials at all. The majority of the results using these methods provided simple kinematics, symmetry assessments and specific joint kinetic information that favoured a heavier prosthetic shank mass.

Mechanical work estimates were only evaluated in four studies including two simulations that found contradicting results with regards to preferable prosthetic mass. The remaining two studies included one that detected no differences in mechanical power under model-defined optimal energy and mass conditions and another that estimated mechanical power only for the purpose of using it as a predictor of metabolic work. It is evident that total estimates of the mechanical work performed by amputees walking wearing different massed prostheses are scarce and those that do exist (Bach et al., 1994), have had trouble in determining the appropriate mass values that may provide a mechanical advantage or disadvantage.

(iii) Metabolic Work

The other major concern when accessing the energy costs of amputees wearing differently massed prostheses is the amount of oxygen consumed (O_2) while walking. Metabolic cost has been an issue of investigation in some of the studies already mentioned as well as others. Skinner and Mote (1989), observed four, old (>59 years), AK subjects walking to steady state with an unweighted limb and with loads of 1.70, 2.74 or 3.97 kg positioned on the prosthesis 17, 25 and 33 cm distal and 7 cm proximal to the knee axis. Results indicated a significant decrease in oxygen consumed with the addition of weight to the prosthetic limb relative to the unweighted limb. The 17 and 25 cm additions of 2.74 and 3.97 kg's proved to be most beneficial. Added mass above the knee drove oxygen consumption up. The study by Bach et al. (1994), revealed a reduction in the oxygen consumption of five, young (mean ± sd = 36.2 ± 6.5 years), AK subjects wearing experimental prostheses loaded for optimal symmetry patterns as predicted by simulation. In all loading situations the oxygen consumption was reduced in comparison to the unweighted limb.

In a study of thirty-nine BK amputees (mean age ± sd = 47 ± 16 years) walking with previously defined "heavy" or "light" prostheses for six to nine minutes, Gailey et al. (1994) reported no significant differences in oxygen consumption when controlling for stump length, age, speed of ambulation and baseline VO_2. Heavy prostheses were denoted by those > 2.27 kg and used by 23 of the 39 participants. Light prostheses, defined as being < 2.27 kg's were used by 16 of the subjects. Lastly, in a recent study, Czerniecki et al. (1994) questioned the metabolic demands of eight, young (33 - 41 years), AK amputees walking at 3 control speeds (0.6, 1.0 and 1.5 m/s), wearing an unweighted prosthesis and a prosthesis with mass additions of 0.68 and 1.34 kg's. It was determined that no significant differences in oxygen consumption exist at any of the speeds studied and for either, unloaded or loaded conditions.

Studies involving the metabolic costs of amputees walking wearing prostheses with different masses seems to be a quick and easy way to evaluate the amputee however, the variance in results currently found in the literature suggests there is still some question as to the appropriate mass and mass distributions to be provided in order to meet the needs of the general amputee population. Despite the above mentioned attempts to test large numbers of amputees, very few conclusions can be generalized from these findings for both AK and BK populations, old and young.

Amputee and Prostheses Inertial Characteristics

In order to ascertain the most accurate estimates of segment masses, center of mass locations and moments of inertia for substitution into mechanical work equations involving amputees, methods different from those used in studies of "normal" human movement must be established. Studies reporting human body segment inertial characteristics based on cadaver data (Dempster, 1955 and Chandler et al., 1975) have been used for investigations of normal locomotion. However, these data are not appropriate for partial segments following amputation or for the attached prosthesis.

To date, only one attempt has been aimed at providing inertial data specifically for fundamental use in amputee studies. Bach et al. (1995) provides a list of inertial characteristics derived from twelve TF prostheses and nine TF residual limbs. It is cautioned that these findings be used only for computer modelling applications and that with the variety of amputation levels and prosthetic componentry available, any "non-modelling" studies should assess prosthetic limb segment parameters on an individual basis. Thus, the "generic" amputee and prostheses data base established by Bach and co-workers was not used however, the methods used to compile results for this data set seemed to be the most simplified, accurate and non-invasive of techniques when compared with others (Contini, 1970; Hanavan, 1964; Jensen, 1978; Solomonidis, 1980; Fernie and Holliday, 1982; Krouskop et al., 1988). The methods of Bach et al. (1995) pertaining to thigh inertial characteristics were based on mathematical models developed by Hatze (1979).

Summary

Following review of the above literature, it is evident that all studies aimed at estimating a prostheses mass that may provide energy benefits for lower extremity amputee gait, do not include evaluations of the existing materials currently used in prosthetic design. "Added mass" trials and simulation studies can mimic the inertial properties and characteristics of existing materials however, they are not directly addressing the question, "Do the lower mass materials (titanium) available to amputees by manufacturers actually provide energy expenditure benefits and if so, what are the mechanical reasons?" In the studies that do exist, the lack of sample size, sample homogeneity, and consistency of findings in mechanical and metabolic measures prompt the need for a large multidisciplinary study to directly address the issues of amputee energetics and prostheses mass. The cited research in this review suggest that it is likely the inertial parameters can be determined and included in biomechanical models of sufficient diagnostic ability to answer these questions.

CHAPTER III - METHODS

Subjects

A total of fifteen unilateral, lower extremity amputees participated in the study. This included two female participants. Seven of the subjects were amputated at the transtibial (TT) level and the remaining eight, at the transfemoral (TF) level. Fourteen subjects (7 TT and 7 TF) were evaluated for measures of mechanical work and metabolic work. One subject (TF) performed the metabolic test only. Table 1 provides a summary of subject descriptive information. All subjects were required to sign an informed consent form following description of the experimental protocol prior to each testing session.

TESTING PROTOCOL

Each subject was required to visit the lab once. Entire sessions lasted 2.5 to 3 hours. Upon entry into the lab, each subject was fitted and aligned with new prosthetic components manufactured from either titanium or stainless steel as determined through random selection prior to testing. New, standardized components were added below the level of the socket. All new components were adapted to the subjects original sockets using an Otto Bock 4-hole socket adapter with the exception of one subject whose original

Table 1: Subject descriptive data.

Subject	Gender	Age (yrs)	Level	Time Since Amputation (yrs)	Cause	Original Components
JR	M	34	TT	9.0	trauma	Graphite
GG	F	63		0.40	p.v.d.	Steel
SH	F	45		2.0	trauma	Titanium
JJ	M	67		2.0	p.v.d.	Steel
BM	M	49		1.60	p.v.d.	Steel
GF	M	37		0.75	trauma	Titanium
RV	M	52		14.0	p.v.d.	Titanium
BL	M	34	TF	2.0	trauma	Graphite
GM	M	57		41.0	cancer	Titanium
RM	M	30		18.0	trauma	Graphite
WA	M	38		2.25	trauma	Titanium
DM	M	58		5.50	p.v.d.	Steel
JS	M	26		12.0	trauma	Carbon/Aluminum
BD	M	73		50.0	trauma	Aluminum/Titanium
JC	M	40		8.0	cancer	Graphite
Mean		46.9		11.24		

socket was duplicated to accomodate the new components. All technical work and clinical fitting and alignment procedures were performed by the same prosthetist to ensure consistency in methods and technique.

Following proper alignment adjustments the subjects were asked to walk across the lab floor with their new prosthesis. Subjects walked under the supervision of a prosthetist until they were comfortable and achieved a gait that felt natural to them. Subjects were given as much time as needed to feel comfortable on the prosthesis. In this manner, subjects were habituated to the new prosthesis.

Alignment adjustments were ongoing during the habituation trials. Following this, subjects then attempted to walk on a motorized treadmill with their new prosthesis. Initial attempts were made at very slow speeds with subjects using the parallel handrails as guides. Gradually, the subjects released the handrails and walked until they could self-select a speed that felt comfortable for them. Again, subjects were given all of the time necessary to feel comfortable on the treadmill with the new prosthesis. This allowed for habituation to the treadmill. All treadmill walking was supervised by a prosthetist in order to adjust for any alignment differences that may exist between treadmill walking and walking on the lab floor.

Measurements of the subject's physical characteristics were then taken. Subject height and mass were measured and recorded using a measuring tape and force plate(AMTI ORG-6), respectively. A heart rate (HR) monitor was strapped around the chest of the subject to track HR activity. Kinematic data were collected by attaching infared emitting diodes (I.R.E.D.'s) on the body of each subject at various joint locations on the following anatomical landmarks: right foot (fifth metatarsophalangeal joint), right heel (lateral side), right ankle (lateral malleolus), right knee (lateral femoral condyle), right hip (greater trochanter), right lumbar 4-5 (inferior edge of the iliac crest vertically aligned with the greater trochanter), right shoulder (inferior to acromion process), right elbow (humeral epicondyle), right wrist, left foot (first metatarsalphalangeal joint), left heel (medial side), left ankle (medial malleolus), left knee (medial femoral condyle), and right ear canal (attached to headpiece). I.R.E.D.'s were connected to two strober units attached to the wastebands of shorts worn by subjects. The strober units were connected to an optoelectronic main processing unit via a central wire (OPTOTRAK/3020, Northern Digital Waterloo, Ontario). The OPTOTRAK/3020 3D motion analysis system employs a 3 axes position sensor to track and identify up to 256 infared markers at a sampling rate of 3500 markers per second. The system is pre-calibrated and has a resolution of 0.01 mm at 2.5 m. It provides a volume of 2 m high by 3.4 m wide at 6 m. The system was supported by a standard IBM compatible computer. Subject segment lengths were recorded with reference to the position of the I.R.E.D.'s. A headpiece, mouthpiece and nose clips required for gas collection were then fitted onto subjects and connected to a metabolic cart via a breathing tube. Subjects were then considered ready for a 4-5 minute walking trial on the treadmill wearing their first experimental prosthesis. Following the first walking trial subjects were given sufficient rest time (as much as requested) and the above procedures were repeated, this time walking on the second experimental prosthesis composed from different materials. Alignments were duplicated by the prosthetist to be identical between trials. Self-selected walking speeds were held constant across walking trials on each of the materials tested.

COMPONENTRY

All subjects, but one, wore their original sockets and were fitted and aligned with titanium and stainless steel componentry from the level of the socket down. Knee and foot components were standardized for all trials. The only differences between the knee and foot components of the titanium and stainless steel prostheses were in their masses. Those subjects with amputations at the TF level walked with the Otto Bock 3R36 titanium knee and the Otto Bock 3R20 stainless steel knee. Both knees were polycentric, constant friction knees with spring-aided extension assists. All subjects were fitted with the Otto Bock 1D10 dynamic foot for testing trials. These componentry arrangements are commonly used by amputees. Where possible, socket and foot adapters and foot bolts made from titanium and stainless steel were interchanged.

Metabolic Data Collection

Before walking on the treadmill subjects stood quietly for 1 minute as oxygen uptake (VO_2) was monitored.

Subjects were connected, through a mouthpiece and low dead space valve, to a custom configured, open circuit system (Ametek /S-3A/I O_2 analyzer, Hewlett Packard 78356A CO_2 analyzer, Ametek R1 flow meter) which was calibrated with gases of known concentration. Expired gas was collected continuously, with VO_2 values recorded at 30 second intervals. In each trial HR was monitored and stored at 15 second intervals, using a Polar Vantage XL Sport tester monitor (Polar Electronics, Finland). An exercise bout length of 4 minutes was chosen to ensure that subjects were at metabolic steady state when data were collected.

Metabolic Cost Determination

Three VO_2 values taken during the last 90 seconds of steady state walking were used to calculate an average estimate of VO_2 for each subject in $l\cdot min^{-1}$. These values were then corrected for body mass and converted to $ml\cdot kg^{-1}\cdot min^{-1}$. Thus, the net submaximal VO_2 for each exercise bout was defined. These data were then converted to $watts\cdot kg^{-1}$ using a respiratory exchange ratio to convert ml oxygen consumed to joules of energy and dividing by 60 seconds. During the same time HR values were recorded at 15 second intervals and averaged to determine a steady state HR for each subject in $b\cdot min^{-1}$.

Determination of Prosthesis and Residual Limb Inertial Characteristics

After walking trials were completed the subject's prostheses were disarticulated below the socket. These components, with shoe on, were then subjected to the following measurements: 1) mass of the prosthesis (via force plate) 2) center of mass location (knife-edge balance) and 3) moment of inertia (pendular suspension). The socket was then removed from the residual limb and subjected to the same measurements.

The residual thigh and shank masses, center of mass locations and moments of inertia were mathematically determined using similar methods and equations as described in Bach et al., (1995) derived from Hatze, (1979). Circumferential measurements at incremental depths, stump length and the vertical distance from the ischial seat to the greater trochanter in TF's, and posterior hamstring flare to the femoral condyle in TT's (cm), were used to produce a geometric model for the estimation of stump characteristics (Bach et al., 1995). Circumferences were measured using a measuring tape at five different locations spanning the external surface of the socket. Socket length was measured with the measuring tape from the apex of the socket vertically, to the midpoint of the ischial seat and, to the patellar tendon bearing surface in TF's and TT's, respectively. The remaining vertical distance to the greater trochanter (Bach et al., 1995) and femoral condyle were also measured.

The stump was then modelled as a stack of n geometric shapes (Bach et al., 1995) where n is the number of incremental depths. The portion of the stump above the ischial seat in TF amputees was modelled as a circular paraboloid (Bach et al., 1995) similar to the elliptic paraboloid used by Hatze (1979). The portion of the stump above the posterior hamstring flare to the femoral condyle in TT amputees was modelled as a circular cylinder. The remaining portions of the stumps in TF and TT amputees were modelled as circular cylinders. Socket and liner thicknesses were taken into consideration when performing external socket measurements and were subtracted to better estimate the stump circumferences. The stump was assumed to have a constant uniform density of 1.05 $g \cdot cm^{-3}$ for TF amputees and 1.09 $g \cdot cm^{-3}$ for TT amputees which are the average densities of the thigh and lower leg cited by Winter (1979).

These measurements were then inserted into equations described by Bach et al., (1995) to arrive at residual limb (stump) inertial characteristics. The stump inertial characteristics were combined with socket inertial characteristics for a single “thigh” or “shank” segment. In TT amputees, the “shank” inertial characteristics were then combined with the inertial characteristics of the remaining componentry below the socket (socket adapter, tube, tube clamp, foot adapter, foot) to determine the mass, center of mass and moment of inertia of a final "shank-foot" segment. An example of one calculation is provided in Appendix I.

Kinematic Data Collection

Kinematic data were collected by sampling the I.R.E.D.'s at 100 frames$\cdot$s^{-1} for 5 seconds during the final 30 seconds of each bout of amputee walking. Optotrak cameras located perpendicular to the plane of motion (sagittal plane) were able to detect and record spacial co-ordinates of the joint centers of subjects.

The resulting Optotrak data was converted to kinematic information using various programs of the biomechanics lab at McMaster University. The total number of 500 samples was reduced to a number which included data points for 1 gait cycle beginning at heel strike of the prosthetic limb. The data were digitally filtered using a Butterworth Low Pass filter with a cutoff frequency of 6 Hz. (Winter et al., 1974).

An 11-segment, sagittal plane model was constructed for both transfemoral (TF) and transtibial (TT), right and left amputees. Model segments included 1 head/neck, 2 upper arms, 2 forearms, trunk, 2 thighs, 1 leg, 1 foot and 1 shank-foot. The shank-foot segment was developed to account for the fixed ankle joint found in components used in this study. Thus, that section of the prosthesis from the knee down was modelled as a single segment. Model thigh and shank-foot segments of the prosthetic limb were adjusted to include the subject's actual inertial parameters determined by the above methods. Inertial characteristics of the remaining model segments were constructed according to Winter (1990). A custom software program calculated joint and segment displacements, velocities (linear and angular) and segmental and system energies.

Mechanical Power Determinations

Work values were computed assuming a negative:positive efficiency ratio of 3:1 (Abbott et al., 1952). Three models were used to calculate mechanical power: 1) based on the kinematics of the center of mass (COM) of the body (Wexch), 2) based on the kinematics of the individual body segments not allowing energy transfers within or between segments (Wn), and 3) based on the kinematics of the segments, allowing transfers of energy within and between segments (Wwb).

The first mechanical power calculation used the kinematics of the COM of the body and assumed complete exchange between potential (PE) and kinetic (KE) energy components (Wexch). PE and KE changes of the body's center of mass at time j were calculated and added to give a total body mechanical energy value (E_j):

$$E_j = PE_j + KE_j = (mgh)_j + \tfrac{1}{2}(mv^2)_j \qquad [1]$$

Where m is body mass (kg), g is acceleration due to gravity ($m \cdot s^{-2}$), h is the vertical position of the center of mass (m) and v is the linear velocity of the center of mass ($m \cdot sec^{-1}$). Total body mechanical work (Wexch) was then calculated for one stride by summing the whole body energy changes using equation [2].

$$Wexch = \sum_{j=1}^{N} \left(\left| E_{j+1} - E_j \right| \right) \qquad [2]$$

Dividing the work values by the time required to complete the stride allowed for the calculation of average mechanical power.

Using a segmental approach mechanical power was also calculated allowing no transfers of energy (Wn) and allowing transfers within and between all segments (Wwb) using equations [3] and [4] respectively.

$$W n = \sum_{i=1}^{s} \sum_{j=1}^{N} \left(\left| \Delta PE_{ij} \right| + \left| \Delta TKE_{ij} \right| + \left| \Delta RKE_{ij} \right| \right) \qquad [3]$$

$$W wb = \sum_{j=1}^{N} \left| \sum_{i=1}^{s} \left(\Delta E_{ij} \right) \right| \qquad [4]$$

PE, TKE and RKE represent the potential, translational kinetic and rotational kinetic energy components, respectively, s is the total number of segments, and N is the number of data points. Energy was calculated for the i^{th} segment at time j, as reported by Pierrynowski et al., (1980) using equation [5].

$$E_{ij} = m_i g h_{ij} + \tfrac{1}{2} m_i v_{ij}^2 + \tfrac{1}{2} I_i \omega_{ij}^2 \qquad [5]$$

where m is the mass of the ith segment (kg), v is its linear velocity ($m \cdot s^{-1}$), ω is its angular velocity ($rad \cdot s^{-1}$), I is its moment of inertia ($kg \cdot m^2$) and h is the vertical position of the center of mass (m). Calculation of energy transfers was achieved by subtracting Wwb from Wn.

Statistical Measures

(i) Metabolic Data

Subjects (n=15)

A one-way analysis of variance (ANOVA) for repeated measures was performed on all subjects with metabolic data to examine the effects of using different component materials. A two factor ANOVA for independent groups (7 TT, 8 TF) was performed [factor 1 = group (TT/TF) and factor 2 = components (Ti./St.)]. In each case, significance was tested for at the $\alpha < 0.05$ level using a two-tailed test.

(ii) Mechanical Data

Subjects (n=14)

A two factor ANOVA on group (TT/TF) and components (Ti./St.) was tested at $\alpha < 0.05$ using a two-tailed test to examine any significant mechanical differences that might exist for these variables.

(iii) Descriptive Statistics

Mean walking velocities, stride rates and stride lengths as well as mean mass differences using titanium and steel were recorded to better interpret the walking characteristics and prostheses inertial differences for each individual. In an attempt to further elaborate on metabolic and mechanical results subjects were divided into categories of age and experience walking on a prosthesis and represented graphically. Lastly, scatterplots of TF and TT prostheses masses against VO_2 and Wwb were arranged to better illustrate the predictive value of prosthesis mass on energy expenditure.

(iv) Subjective Assessments

Following the entire testing session subjects were asked if they preferred one prostheses over the other and if they noticed the mass differences betweeen the two.

CHAPTER IV - RESULTS AND DISCUSSION

Prostheses Inertial Characteristics

Table 2 presents the mass, moment of inertia (I) and COM differences of prostheses assembled from titanium and stainless steel for all subjects. As expected, all prostheses assembled from stainless steel weighed more and had segments with moment of inertia and COM (distance from the proximal joint) values greater than or equal to those assembled from titanium. Consequently, subjects had a greater mass wearing stainless steel as opposed to titanium components. Subjects with amputations at the transtibial level had a considerably smaller mean mass difference (183 grams) than transfemoral subjects (351 grams). These figures are in line with those reported in orthopedic industry product catalogues (Blohmke, 1989).

Walking Characteristics

As self-selected walking velocities remained constant for titanium and steel trials, stride rates and lengths changed due to the differently massed components (table 3). As expected, when stride rates were slower strides lengthened and when rates were faster, strides shortened. This relationship however, expressed no

Table 2. Subject masses and prostheses inertial characteristics.

Level	SS	Subject Mass (kg)		Mass Diff. (g)	Moment of Inertia (kg/m²)				Center of Mass (m)			
					Thigh		Shank-foot		Thigh		Shank-foot	
		Ti	St		Ti	St	Ti	St	Ti	St	Ti	St
TT	JR	82.97	83.15	180	-	-	.037	.047	-	-	.164	.164
	GG	80.43	80.72	290	-	-	.064	.070	-	-	.170	.168
	SH	70.61	70.76	150	-	-	.026	.038	-	-	.145	.144
	JJ	70.13	70.34	210	-	-	.054	.073	-	-	.126	.126
	BM	100.81	101.02	210	-	-	.065	.086	-	-	.154	.154
	GF	97.67	97.81	140	-	-	.069	.076	-	-	.159	.159
	RV	73.71	73.81	100	-	-	.019	.031	-	-	.160	.160
			Mean	182.86								
TF	BL	74.40	74.70	300	.077	.077	.039	.137	.18	.18	.360	.330
	GM	72.21	72.68	470	.093	.093	.073	.121	.20	.20	.350	.310
	RM	84.14	84.52	380	.107	.107	.020	.098	.18	.18	.340	.315
	WA	71.49	71.78	290	.136	.136	.012	.110	.23	.23	.285	.242
*	DM	69.72	70.02	300	-	-	-	-	-	-	-	-
	JS	70.09	70.39	300	.079	.079	.074	.122	.21	.21	.365	.335
	BD	70.46	70.93	470	.101	.101	.074	.121	.26	.26	.310	.290
	JC	102.66	102.96	300	.038	.038	.056	.084	.20	.20	.315	.285
			Mean	351.25								

* *Metabolic data only.*

particular pattern for titanium and steel and specifically did not show a pattern that would favour the use of a heavier material over a lighter one as suggested by some authors (Mena et al. 1981; Menkveld et al., 1981; Tashman et al., 1985). Rather, it would agree with Godfrey et al., (1977) who observed no differences in pace rate with the addition of 113.4 and 226.8 gram weights attached to the prosthetic feet of six AK amputees.

Table 3. Subject walking characteristics.

Level	*Subject*	*Walking Velocity (m/s)*	*Stride Rate (stride/s)*		*Stride Length (m)*	
			Ti	St	Ti	St
TT	JR	0.91	0.78	0.80	1.16	1.14
	GG	0.68	0.72	0.86	0.95	0.79
	SH	0.86	0.88	0.79	0.97	1.09
	JJ	0.64	0.69	0.77	0.92	0.83
	BM	0.77	0.85	0.82	0.90	0.94
	GF	0.95	0.81	0.78	1.18	1.22
	RV	1.14	0.83	0.94	1.38	1.21
	Mean	0.85	0.79	0.82	1.07	1.03
TF	BL	0.82	0.73	0.73	1.12	1.12
	GM	0.82	0.69	0.67	1.18	1.23
	RM	0.86	0.71	0.71	1.20	1.20
	WA	0.64	0.67	0.67	0.96	0.96
	JS	0.91	0.74	0.79	1.24	1.16
	BD	0.64	0.85	0.84	0.76	0.76
	JC	0.64	0.69	0.75	0.92	0.86
	Mean	0.76	0.73	0.74	1.05	1.04

As speeds were held constant across trials and stride rates (TT/Ti mean = .79 stride/s, TT/St mean = .82 stride/s; TF/Ti mean = .73 stride/s, TF/St mean = .74 stride/s) and lengths (TT/Ti mean = 1.07 m, TT/St mean = 1.03 m; TF/Ti mean = 1.05 m, TF/St mean = 1.04 m) differed only slightly, it seems that these variables had little effect on mechanical and metabolic findings (see next section). The differences in mean walking velocities of TF and TT subjects may have affected oxygen consumption values. Even though TT subjects walked faster on average the oxygen consumed was less, (fig. 1) possibly because they were walking at speeds closer to their optimum oxygen consumption pace.

Energetics

(i) Metabolic Work

Metabolic work estimated from oxygen consumption was greater for the stainless steel than the titanium trials but not significantly [$(F1,14)= 1.45$, $p<.249$] when examining all 15 subjects (fig. 1). As well, TF and TT subjects experienced no significant metabolic effects due to on the different materials [$(F1,13)= 1.31$, $p<.274$] however, TF subjects consumed significantly more oxygen than TT subjects [$(F1,13)= 11.34$, $p<.005$] as found in earlier work (Waters et al., 1978). This indicates that the level of amputation is more important than the material used when considering

AMPUTEE OXYGEN CONSUMPTION: EFFECT OF MATERIALS

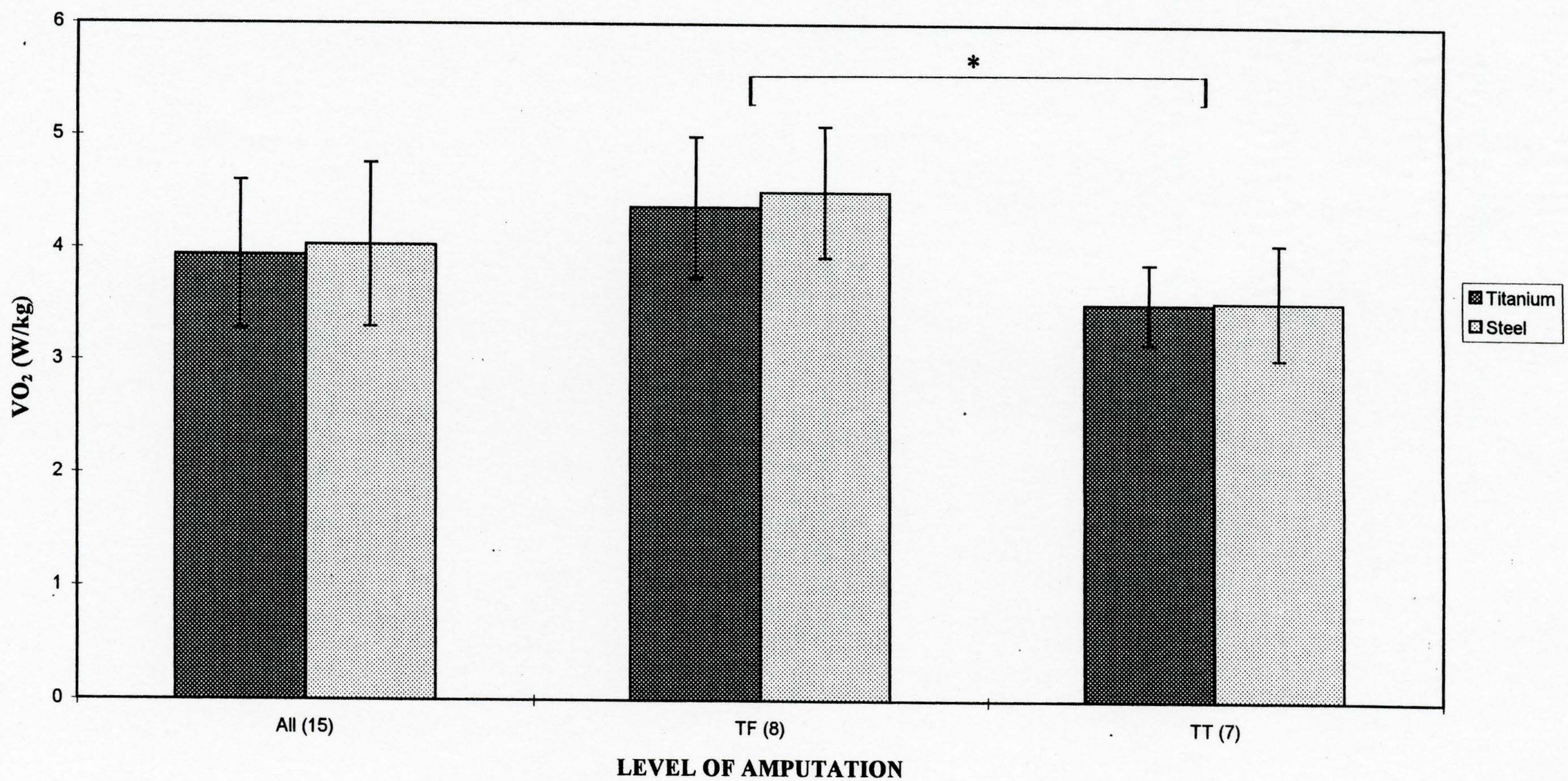

Figure 1. Effects of titanium and stainless steel on amputee metabolic work (W/kg). * p < .05.

metabolic work.

(ii) Mechanical Work

Total mechanical work measurements on all 14 subjects were significantly reduced [$(F1,12) = 4.85$, $p<.048$)] when walking with titanium components as compared to stainless steel as hypothesized (fig. 2). Differences between TF and TT subjects (fig. 2) were small and not significant [$(F1,12) = .013$, $p<.912$].

These results indicate that although a reduction in prostheses masses evoked a reduction in mechanical work, the decrease in mechanical work was not accompanied by a significant decrease in oxygen consumed, as would be expected. The rationale for these findings begins with questioning the relationship between mechanical and metabolic estimates and the reciprocal ability or disability of one to explain the other. In this study, one explaination may be that the amount of mechanical work required for walking between the two materials is not great enough to be detected in the physiological measurements thereby enforcing the notion that the relationship between mechanics and metabolics is not a direct one. This "window" or ratio which defines the point where one measure may respond to the other is likely to be where the misunderstanding lies and is probably responsible for results found in similar studies reflecting low correlations

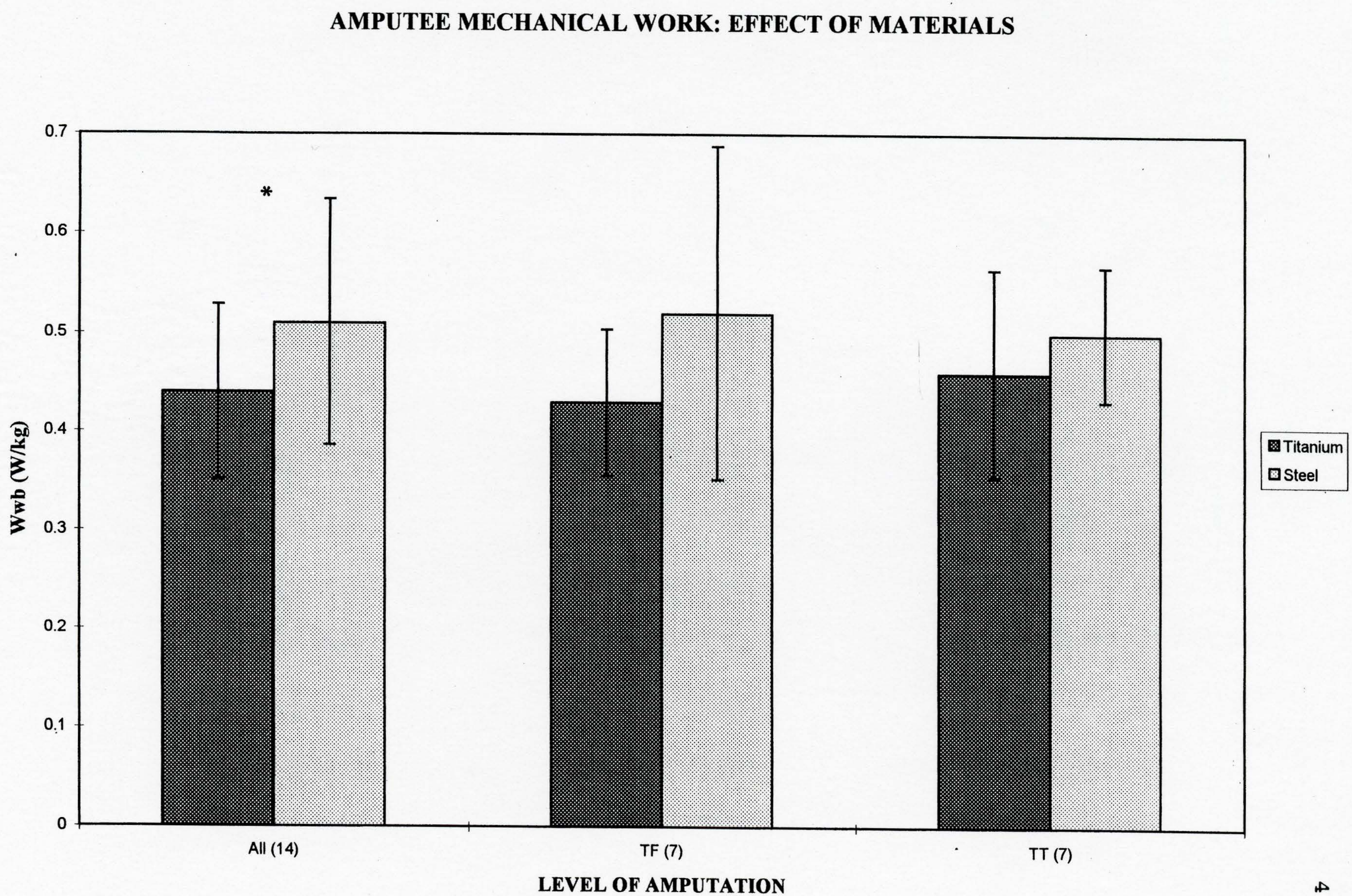

Figure 2. Effects of titanium and stainless steel on amputee mechanical work (W/kg). * $p < .05$.

of mechanical and metabolic measures (Skinner and Mote, 1989; Bach et al., 1994; Beck and Czerniecki, 1994).

At first glance, these findings re-emphasize the caution expressed by several authors in correlating the two different measures of work (Burdett et al., 1983; Williams, 1985; Daniels, 1985 and Fetters et al., 1990). These concerns are based on the inability of biomechanical models to account for isometric work, non-sagittal plane motion, cocontraction of antagonistic muscles, passive tissue contributions and utilization of stored elastic energy. However, current models have been able to provide good correlations regardless (Dowling et al., 1995).

The above results should be interpreted in two ways. Firstly, mechanical and metabolic findings should be viewed independently. These subjects were mechanically walking differently on the two materials even though it was not reflected in their stride rates and lengths. It could have been that the trunk and upper extremities were responsible for these differences. This explanation was explored by examining the work done by the lower limbs. The mechanical work was divided into that which was being done by the sound limb and the prosthetic limb in anticipation that the differences would be seen in the lower limbs. The values however, suggested no sagittal plane differences for titanium or steel within each leg, nor between both legs for each condition. Thus, it did appear that the head, arms and trunk segments (HAT) were where the differences existed if non-sagittal plane (frontal) differences are excluded. The data for the HAT segments solidified this notion and seemed more relevant than any differences in non-sagittal plane mechanics. This led to the use of the mechanical power equation which allowed transfers with the trunk and the legs (Wwb). As well, the mass and inertial differences between materials may have been great enough to change the mechanics significantly. Metabolically we know the subjects were unaffected thus the workload that had to be generated to walk on both materials was no different.

Secondly, if you combine both measures after viewing them independently it might be concluded that the metabolic changes were time and workload dependent and that some subjects were not working long enough to realize the differences whereas the mechanical changes are not time dependent.

Alternative explanations for the results can be provided when considering the population under investigation. Since little time has been devoted to a comprehensive examination of amputee energetics wearing prostheses made from different materials, future studies may include working backwards to analyze the walking kinematics of these individuals. What could be happening is that the differences in the inertial characteristics of the differently massed materials may be compensated for by the mechanisms built into the prosthesis itself. Mainly in TF subjects slight alterations in the spring assist of the knee joint may relieve natural tissues of the body of accounting for the increased energy necessary to achieve the same speed, stride rate and stride length (no extra muscular work would have to be done). Since the current science of clinical prosthetics involves making purely subjective alignment decisions by the prosthetist the best possible method of control for alignment differences between trials would be to have the same prosthetist following the same philosophies and techniques in each adjustment. In this study a great deal of time was devoted to attaining an exact duplicate when assembling and aligning the differnt prosthetic limbs. Thus, to exclude these factors as possible causes for the outcome of results great confidence must be taken in the prosthetist making adjustments in order to reduce human error to a minimum. Future studies may include monitoring any small kinematic differences in the prosthetic segments of these individuals. Also, performing alignments with the use of specialized alignment jigs may be a consideration. Lastly, even though this subject sample does provide a large representation of the population perhaps larger numbers of selected individuals can help to better detect the effects of individual differences amongst subjects.

The results of this study and that of Bach et al. (1994) who found the reverse (metabolic differences and no mechanical differences) should be understood with some knowledge of the limitations when providing biomechanical rationale for physiological responses. The results nonetheless, are valuable to understand what is happening to the amputee in both domains under experimental conditions and may lead to better understandings of the connection between the two disciplines. As well, this study and the study by Bach et al. (1994), represent the first of more energetic studies in the area of amputee gait and with further analyses a better understanding of how well adapted the current models and techniques are to amputee populations may result.

Effects of Materials, Age and Length of Time on a Prosthesis

(i) Metabolic Work

In an attempt to better understand the individual differences that exist between the subjects in this study, the interaction of materials with age and, length of time on a prosthesis have been graphically represented. Figure 3 represents oxygen consumption values for all eight TF subjects and also includes values for TF subjects

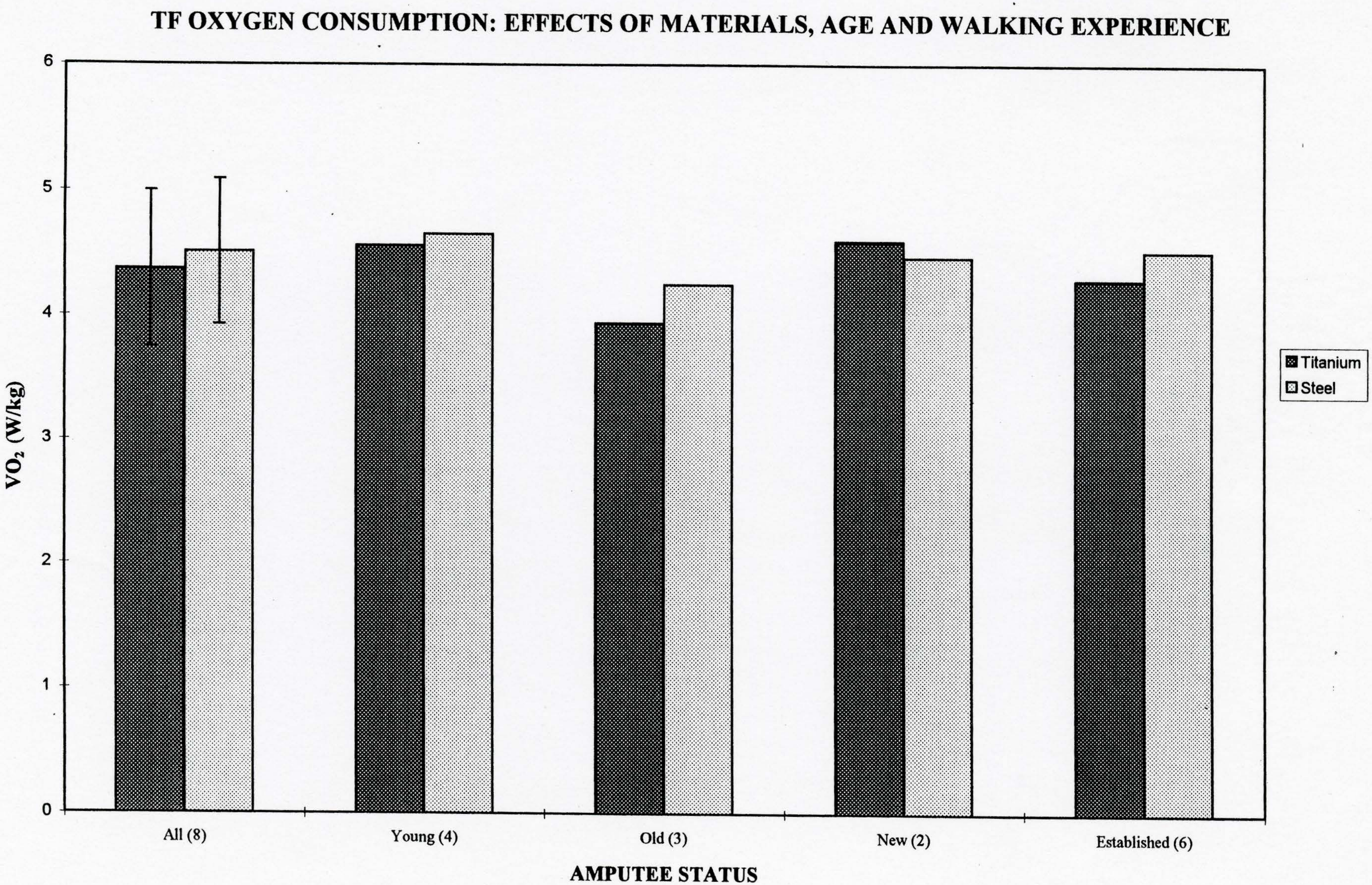

Figure 3. Effects of materials, age and experience walking on a prosthesis on TF oxygen consumption (W/kg).

categorized by age (young(4), < 40 yrs.; old(3), > 50 yrs) and experience walking on a prosthesis (new(2), ≤ 2.25 yrs; established(6), > 2.25 yrs). Immediately, it is apparent that the older subjects consumed less oxygen than younger subjects however, this is probably a result of these subjects walking at slower speeds. Thus, it is important to interpret this graph knowing that speeds will have differed between subjects and groups but not between materials. Also, selecting subjects to meet the criteria of each group resulted in varying numbers of subjects for each category therefore, descriptions will only be qualitative and limited mostly, to within group comparisons.

In all categories except the new walkers oxygen consumption was less for titanium. Perhaps the control mechanisms of new walkers have a harder time adapting to decreased inertial changes even though these particular subjects have original componentry already assembled from lightweight materials (table 1). It appears that older and more established TF walkers benefit the most from the use of titanium. This would be expected as a lightweight prosthesis may better accommodate the loss of musculature that is associated with old age and is normally used to drive the prosthesis forward. For example, the same amount of musculature does not work as hard when initiating swing phase of the prothetic limb using the lighter prosthesis.

Also, an established walker (not always an older subject) may have better control systems that could be more sensitive to small inertial changes due to their experience over the years wearing different prostheses with different components as well as different shoes. This may lead to a more economical recruitment of muscles to be used in the walking movement. The combination of being an older and more experienced walker may realize the greatest benefits from lightweight componentry at least in the case of TF amputees.

Figure 4 represents oxygen consumption values for all seven TT subjects and also includes values for TT subjects categorized by age (young(2), < 40 yrs.; old(3), > 50 yrs.) and experience walking on a prosthesis (new(5), ≤ 2 yrs.; established(2), > 2 yrs.). Contrary to TF results, in this case the new TT walkers achieved the lowest oxygen consumption values relative to other TT categories but again, comparisons using these values do not reflect constant walking speeds. There does however, seem to be metabolic differences that exist between new TF and new TT subjects. The new TT subjects consumed much less oxygen as would be expected being that it is more difficult to walk without the natural knee joint.

TT OXYGEN CONSUMPTION: EFFECTS OF MATERIALS, AGE AND WALKING EXPERIENCE

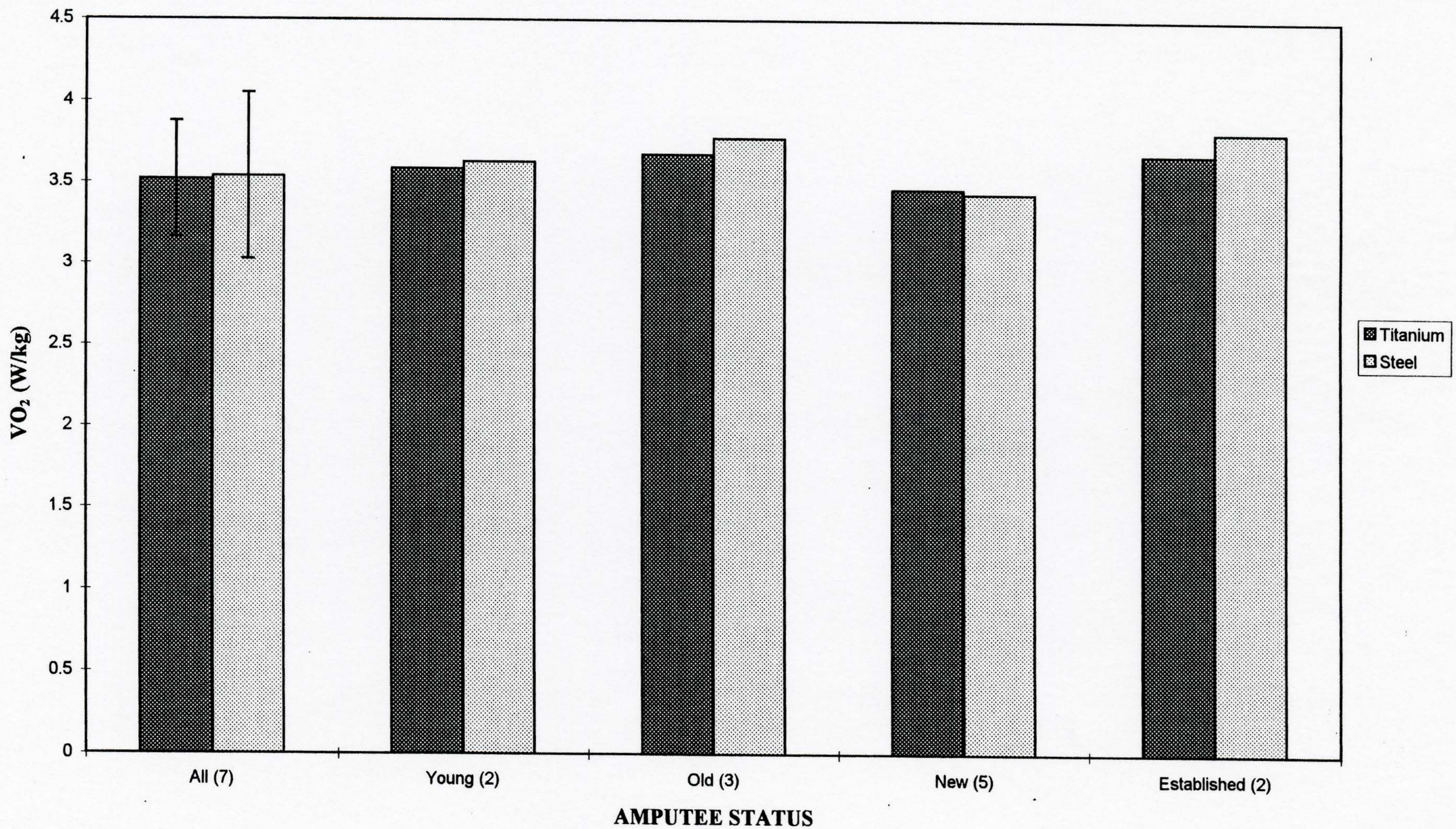

Figure 4. Effects of materials, age and experience walking on a prosthesis on TT oxygen consumption (W/kg).

One general finding similar to the TF results for materials, age and walking experience was that in every category except for new walkers the use of titanium prompted less oxygen consumption. Once again at least when subjects are arranged categorically, it appears as though the older and more established TT walkers realize the greatest benefits from lighter materials. This is probably due in part to similar reasons expressed for TF subjects.

(ii) Mechanical Work

Figure 5 represents mechanical work values for all seven TF subjects and when categorized by age (young(4), < 40 yrs.; old(2), > 50 yrs.) and by experience on a prosthesis (new(2), ≤ 2.25 yrs.; established(5), > 2.25 yrs.). It appears that both young and old groups are affected similarly by changing materials as both display results favouring titanium that are close in magnitude. Established TF walkers also benefit from titanium but these differences are less pronounced. New walkers appear to be generating nearly the same mechanical power regardless of what material is being used. Once again, as was apparent in oxygen consumption results for materials, age and walking experience, it was evident that new walkers were the only group that did not produce lower work values for titanium.

Mechanically speaking, the reasons for lesser

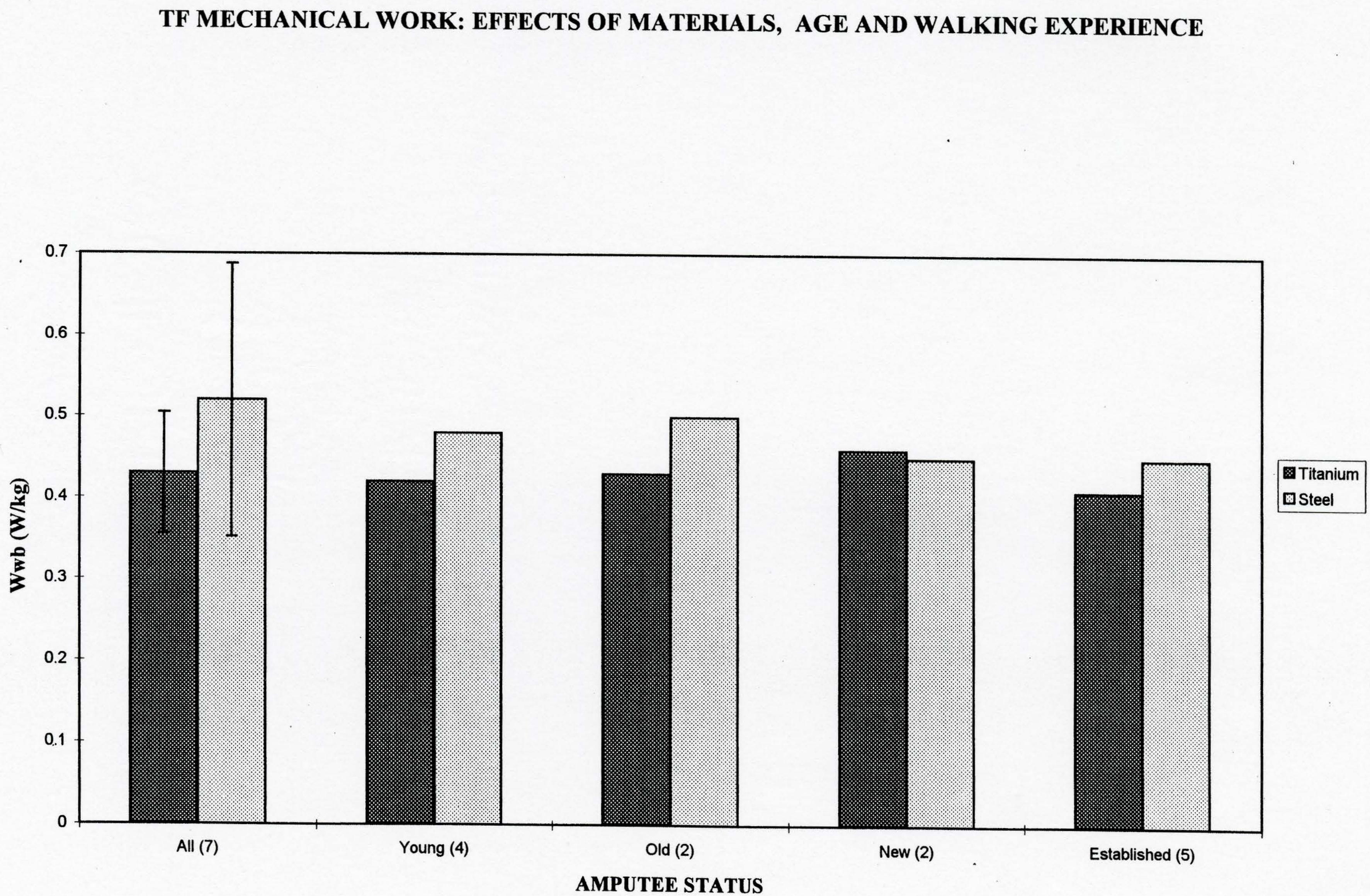

Figure 5.. Effects of materials, age and experience walking on a prosthesis on TF mechanical power (W/kg).

differences in new TF subjects may lead back to the issues of control and adaptation. How long does it take for an individual to neurologically (not to mention psychologically) adapt to the loss of a limb especially when the removal of a major locomotor joint must be artificially replaced? This suggests an index of sensitivity that increases with experience and may not be developed enough yet to detect and adapt for the small differences in material mass, thereby resulting in less economical movement.

Lastly, Figure 6 illustrates mechanical work values for all seven TT subjects and when categorized by age (young(2), < 40 yrs.; old(3), > 50 yrs.) and by experience on a prosthesis (new(5), ≤ 2 yrs.; established(2), > 2 yrs.). In all cases it was apparent that the mechanical work done on titanium was less than stainless steel including that done by new walkers however, only by a narrow margin for this category. Young and established TT subjects seem to receive the greatest mechanical benefits using the titanium prostheses. Again, probably relating to explanations described above.

To summarize the combined effects of materials, age and experience walking with a prosthesis on metabolic and mechanical work, it was apparent that relatively new

TT MECHANICAL WORK: EFFECTS OF MATERIALS, AGE AND WALKING EXPERIENCE

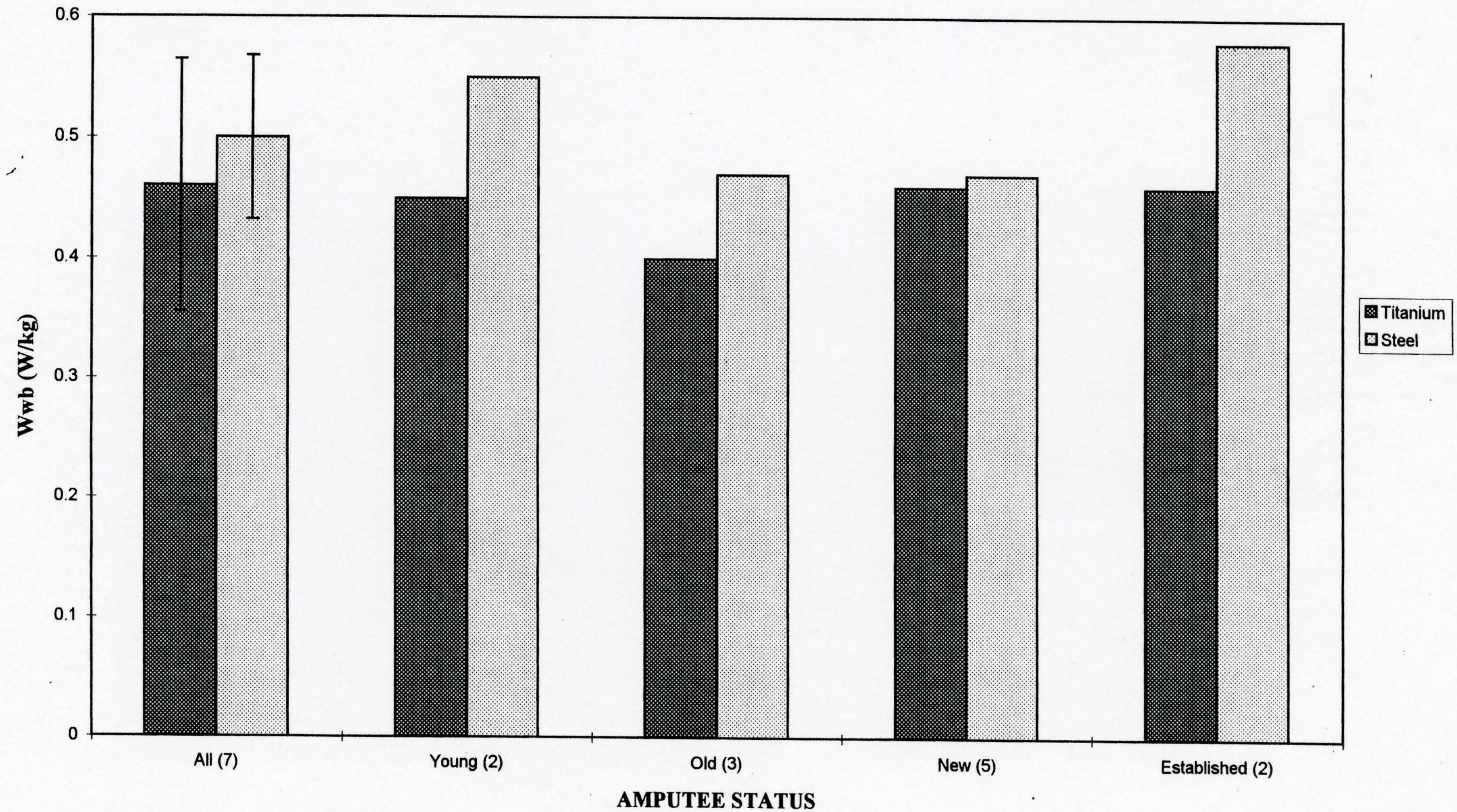

Figure 6. Effects of materials, age and experience walking on a prosthesis on TT mechanical power (W/kg).

amputees amputees did not receive the benefits experienced by other categories of subjects in this study. It seems that length of time walking on a prosthesis may be a more valuable indicator of how an amputee might perform on differently massed materials in comparison to the age of the individual. Although differences were small, and the analysis was limited to being descriptive only, older and more established walkers appeared to perform better mechanically and metabolically walking on titanium. In this study it was evident that the differences in prosthesis mass effected mechanical economy slightly more than metabolic economy when age and experience walking on a prosthesis are considered.

Prediction of Energy Expenditure Using Prostheses Mass

(i) Metabolic Work

Figures 7 and 8 illustrate the relationship between prosthesis mass and VO_2 for subjects in this study. It is apparent that for TF subjects, mass of the prosthesis was not a very strong predictor of VO_2 ($r = .04$) as some individuals experienced their lowest VO_2 scores at the highest prostheses mass values. This relationship appears to be much stronger for transtibial subjects ($r = .61$), however, it is evident that the greatest prostheses masses do not equate to the highest VO_2 values and re-emphasize the lack of increasing mass effects on oxygen consumption for

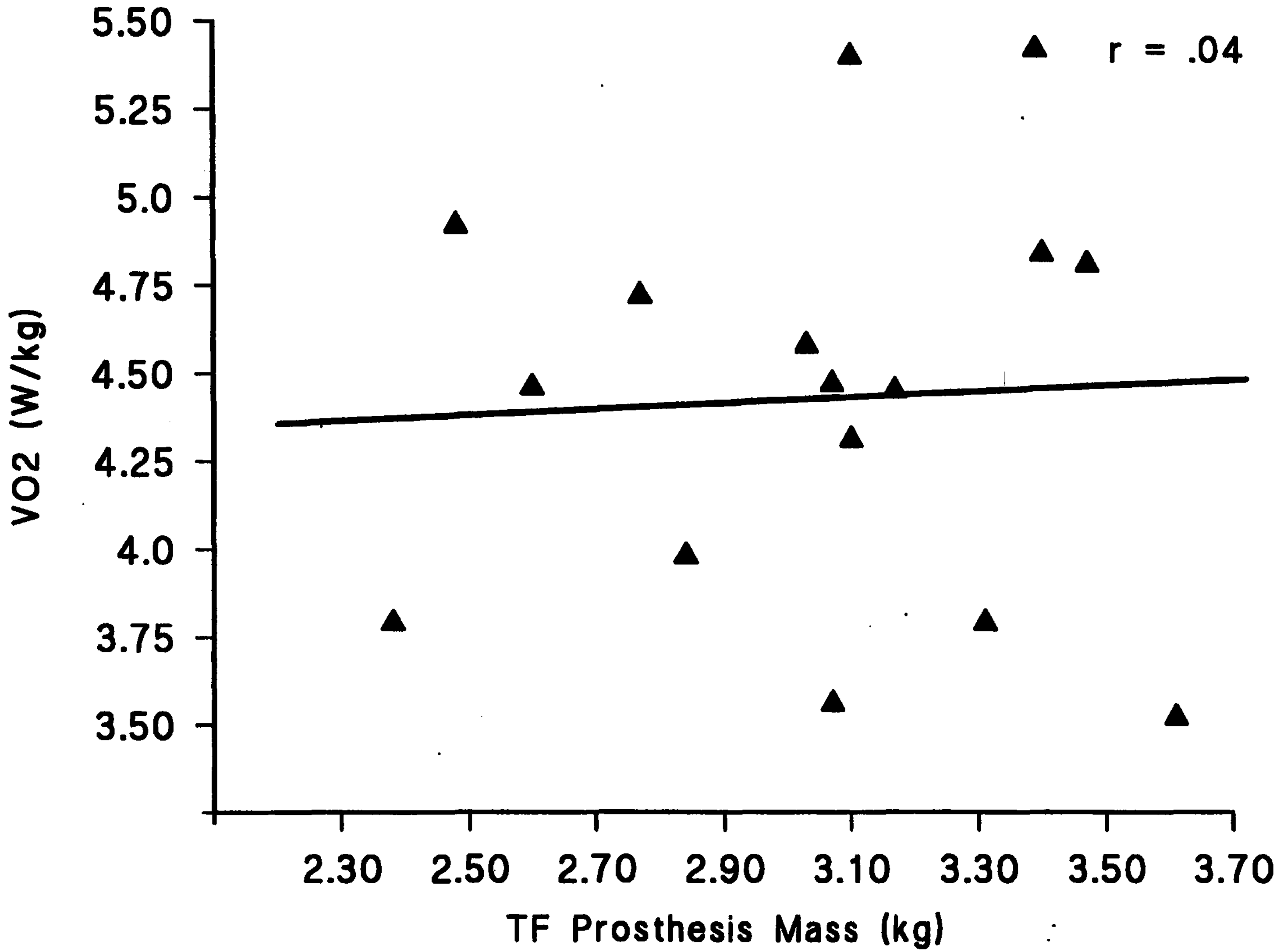

Figure 7. The relationship of oxygen consumed (W/kg) and TF prosthesis mass (kg).

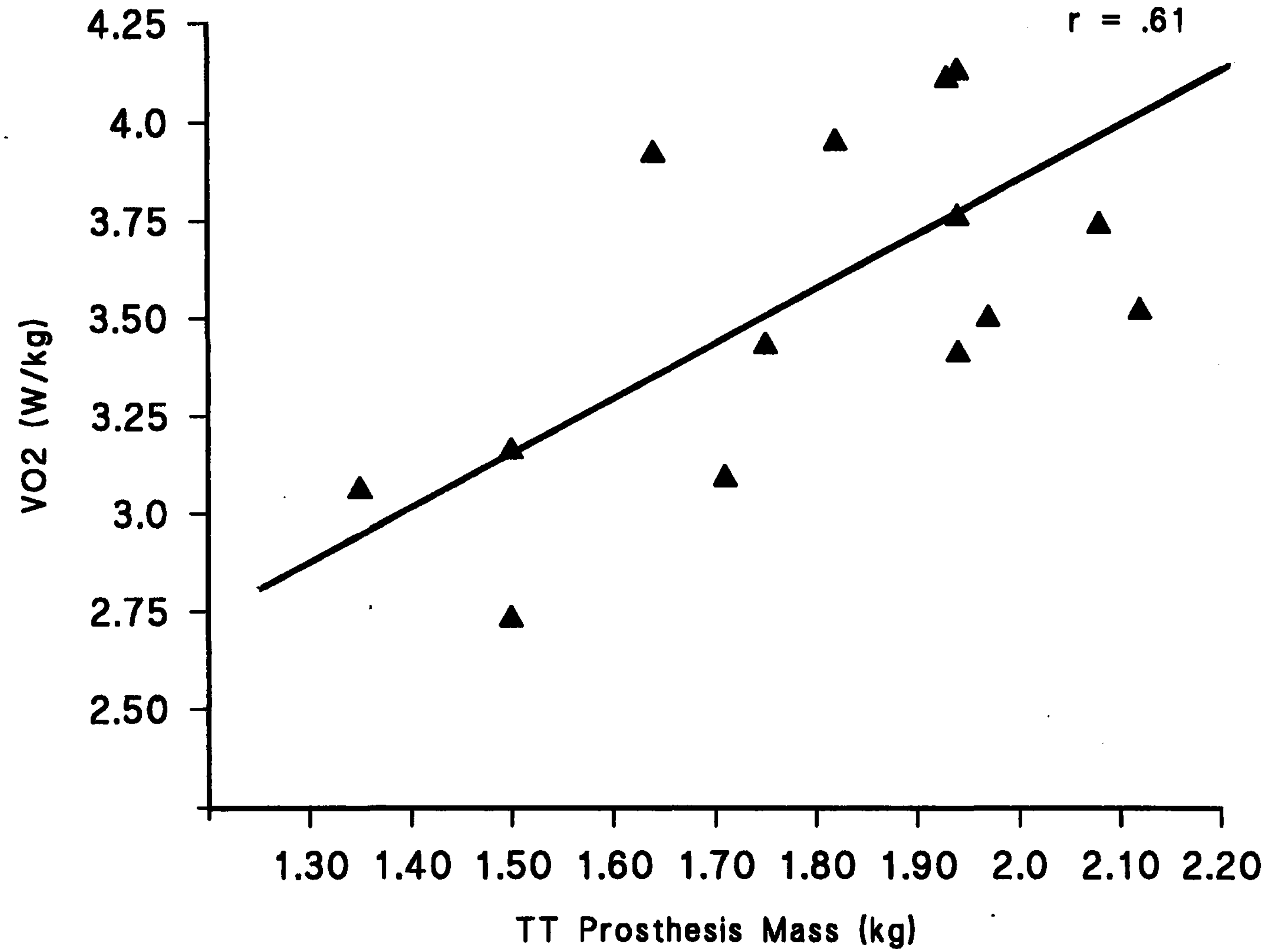

Figure 8. The relationship of oxygen consumed (W/kg) and TT prosthesis mass (kg).

both TF and TT amputees as observed by other authors (Gailey et al., 1994; Czerniecki et al., 1994).

This data lends support to the idea of an "energy" optimum prostheses mass that is individually defined (Bach et al., 1994). The above description of the effects of age and experience walking on a prosthesis on VO_2 can be used to further breakdown the interrelated effects of these variables with mass (different materials) and their combined ability for predicting VO_2.

(ii) Mechanical Work

Figures 9 and 10 illustrate the prosthesis mass-mechanical work (Wwb) relationship for subjects in this study. Both TF and TT prostheses are not very good predictors of Wwb ($r = .09$ and $r = .13$, respectively) displaying various inconsistencies in values. As was the case for TF mass values and VO_2, some (but not all) individuals do the least amount of work at higher mass values. TT prostheses masses do not display any particular predictive pattern as most Wwb values are constant across differing masses.

The idea of an optimal energy mass may also apply to Wwb values. However, when Bach et al. (1994) attempted to provide these optimums via computer simulation, the results were unsuccessful in reducing total mechanical power.

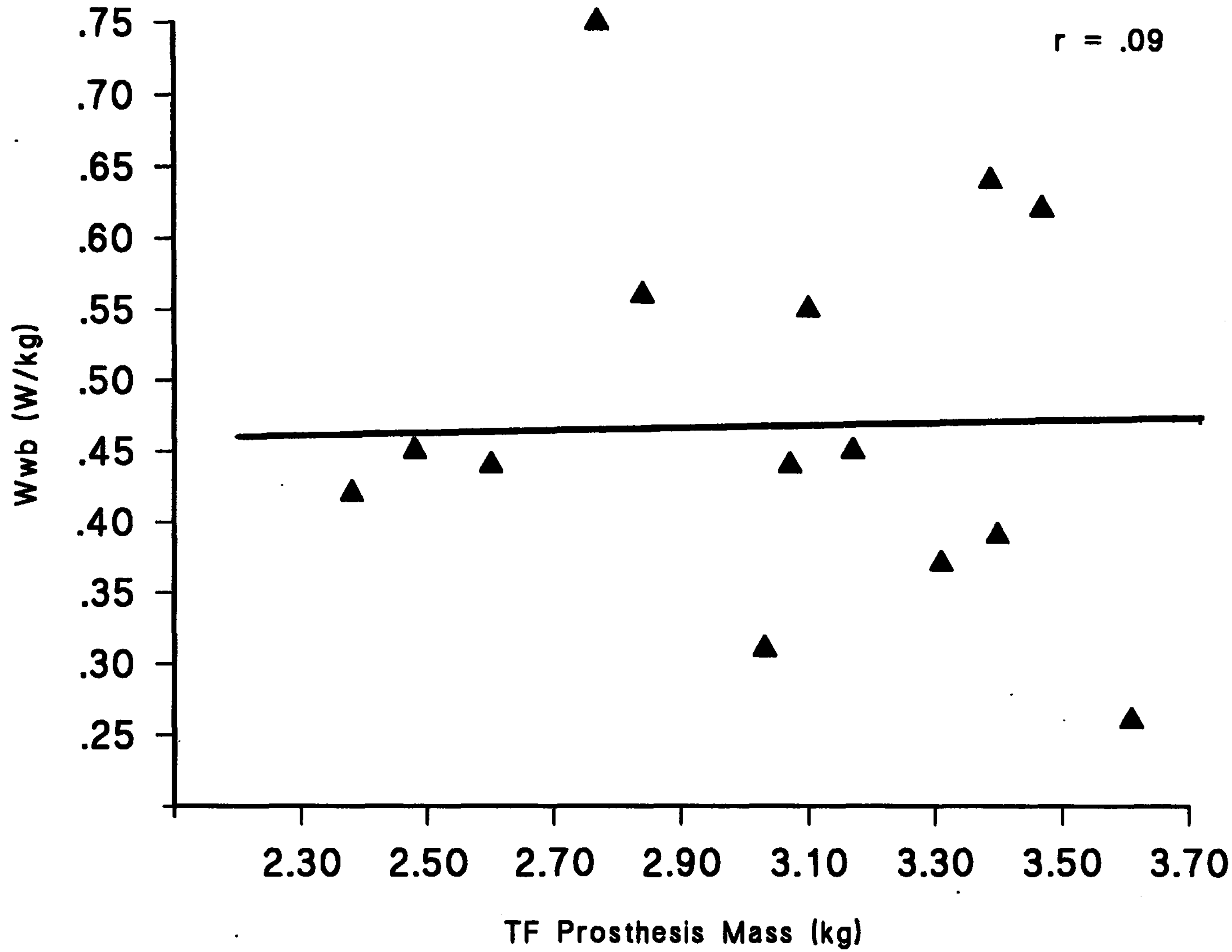

Figure 9. The relationship between mechanical power (W/kg) and TF prosthesis mass (kg).

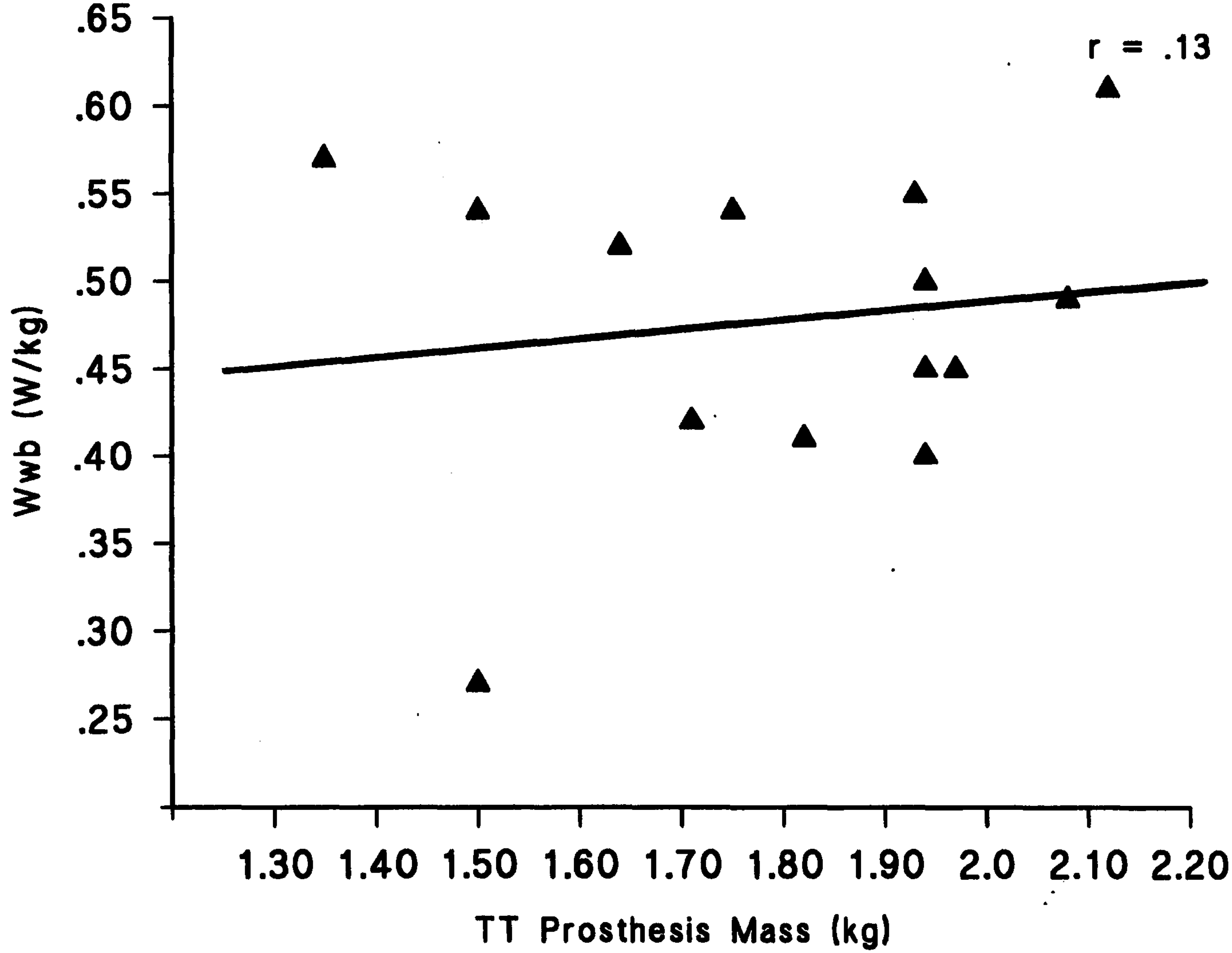

Figure 10. The relationship between mechanical power (W/kg) and TT prosthesis mass (kg).

Subjective Assessments

As the subjects walked they were asked for their own subjective choice of which prosthesis they felt more comfortable on. Replies varied greatly with approximately half of the group unknowingly choosing titanium and half choosing steel. Three individuals felt that the titanium prosthesis was heavier despite mass measurements that proved them wrong. Two subjects felt that steel was lighter than titanium and the remaining subjects afterwards said that they could detect the correct mass differences without difficulty.

CHAPTER V - CONCLUSIONS AND RECOMMENDATIONS

Amputees that walk wearing titanium and stainless steel prostheses have different masses which affects their metabolic and mechanical work output. The results of this study have indicated that the metabolic costs of amputees walking on titanium and stainless steel components exhibited small, non-significant changes whereas, the mechanical work required to walk was significantly different between the two materials. Also, the inertial characteristics of prostheses and walking characteristics of amputees were different when components were assembled from the different materials.

The energy costs of subjects with transtibial and transfemoral amputations were not significantly affected by the choice of materials even though mechanical differences did exist. However, small numbers of subjects divided by age and walking experience on a prosthesis did give an indication that the choice of titanium over steel may be to their benefit both metabolically and mechanically. The categories least effected by the difference in materials were new amputees and those gaining the greatest savings were the older and more established walkers. Further study with more homogeneous samples involving these characteristics as well as others (eg. cause of amputation) with lightweight componentry seems to be a worthy pursuit. One limitation to such a pursuit experienced in this study is that it is apparent that the amputees that would benefit the most from lightweight componentry have difficulty in treadmill walking without the use of an aid and are not capable of attaining a steady state metabolically. The current measurement and testing protocols would have to be adapted to handle these realities.

As a reply to industrial concerns regarding the continued distribution of both materials it would be recommended that both titanium and stainless steel continue to be offered. Individual differences among clients (clinicians and patients) will always remain and even though titanium did not display all of the expected economical benefits for subjects, the mechanical properties of its use in walking do appear to be superior. In addition, the use of titanium for amputees that participate in more active exercise may be beneficial for those with such lifestyles. However, it should be noted that industrial statements regarding the expected economical benefits of their products should be more carefully examined and researched, especially when dealing with the effectiveness of lightweight materials.

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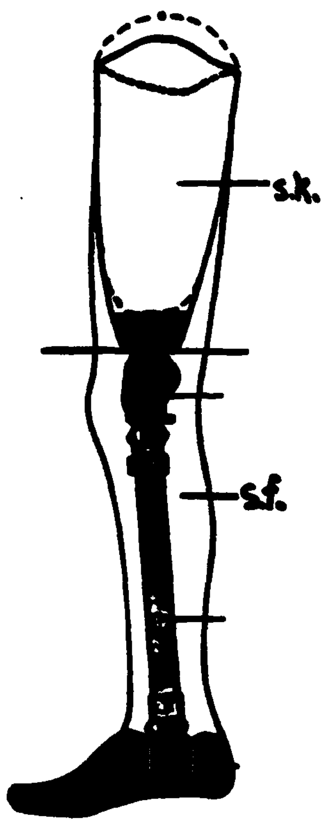
Appendix I

Calculation of Prosthesis and Residual Limb Inertial Characteristics

APPENDIX I

Calculation of Prosthesis and Residual Limb Inertial Characteristics

Prosthesis



<u>Socket Measurements</u>:

-> mass (force plate)
center of mass (knife edge balance)
moment of inertia (pendular suspension)

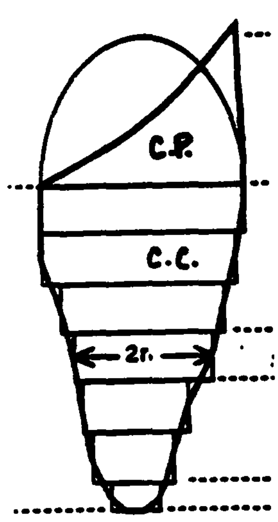
		TF	TT
Example: mass (M_{sk})	=	*1.27kg*	*.51kg*
center of mass (X_{sk})	=	*.24m*	*.215kg*
moment of inertia (I_{sk})	=	*.322kg/m^2*	*.149kg*
moment of inertia(Icmsk)	=	*.017 kg/m^2*	*.0057kg/m^2*

<u>Componentry Measurements</u>: (below sk.)

-> mass (force plate)
center of mass (knife edge balance)
moment of inertia (pendular suspension)

	TF Ti/St	TT Ti/St
Example: mass (M_{sf})	*=1.82 kg / 2.12 kg*	*1.43kg/1.57kg*
center of mass (X_{sf})	*=.36 m / .33 m*	*.24m/.25m*
moment of inertia (I_{sf})	*=.503/ .595 kg/m^2*	*.44/.607kg/m^2*
moment of inertia(I_{sfcm})	*=.039/.137 kg/m^2*	*.035/.042kg/m^2*

Residual Limb



<u>Stump Measurements</u>:
(Bach, 1994; Hatze, 1979)

	Circular Parabaloid	Circular Cylinder
Mass ->	$C = 2\pi r$	$C = 2\pi r$
	$V = {}_{1/2}\pi r^2 h$	$V = \pi r^2 h$
	$^*m_i = DV$	$m_i = DV$

* where $D = 1.09\ g\cdot cm^{-3}$ (TT)
$D = 1.05\ g\cdot cm^{-3}$ (TF)

COM ->	$^*d_i = {}_{2/3}h$	$d_i = {}_{1/2}h$

* where h = height (m) of each parabolic or cylindrical section

	Circular Parabaloid	Circular Cylinder
Moment of Inertia -> I_{gtst}	$= {}_{1/18}m_i(3_i r^2+h_i^2)$	$= {}_{1/12}m_i(r_i^2+h_i)$

TOTALS:

Mass $= \sum_{i=1}^{n} m_i$

Center of Mass $= \sum_{i=1}^{n} m_i d_i / m_{st}$

Moment of Inertia

TF $I_{tgt} = \sum I_{gtst}(CP) + \sum I_{gtst}(CC)$

TT $I_{tfc} = \sum I_{fcst}(CC)$

Stump Measurement Example: (Bach, 1994; Hatze, 1979) - One Section

TF Prosthesis (Circular Parabaloid)	*TT Prosthesis (Circular Cylinder)*
$C = 2\pi r$	$C = 2\pi r$
$.595 = 2\pi r$	$.463 = 2\pi r$
$r = .095m$	$r = .074m$
$V = {}_{1/2}\pi r^2 h$	$V = \pi r^2 h$
$V = {}_{1/2}\pi(.095m)^2(.11m)$	$V = \pi(.074m)^2(.125m)$
$V = .00155m^3$	$V = .00215m^3$
$V = 1.55l$	$V = 2.15l$
$m_i = DV$	$m_i = DV$
$m_i = 1.05\ kg/l \times 1.55l$	$m_i = 1.09kg/l \times 2.15l$
$m_i = 1.63\ kg$	$m_i = 2.34\ kg$
$d_i = {}_{2/3}h$	$d_i = {}_{1/2}h$
$d_i = {}_{2/3}(.11m)$	$d_i = {}_{1/2}(.125m)$
$d_i = .073\ m$	$d_i = .0625\ m$

** remaining sections would be calculated as circular cylinders*

$$Mass = \sum_{j=1}^{N} m_i$$
$$= 5.18\ kg$$

$$COM = \sum_{j=1}^{N} m_i d_i / m_{st}$$
$$= \frac{.904\ m/kg}{5.18\ kg}$$
$$= .17\ m$$

$$I_{cpst} = {}_{1/18} m_i(3_i r^2 + h_i^2)$$
$$= 1/18(1.63)[3(.095)^2 + (.11)^2]$$
$$= .0057\ kg/m^2$$
(One CP section only)

$$I_{gtst} = I_{cpst} + d_i^2(m_i)$$
$$= .0057\ kg/m^2 + .073m^2(1.63kg)$$
$$= .014\ kg/m^2$$
(One section only)

$$I_{tgt} = \sum I_{gtst}(CP) + \sum I_{gtst}(CC)$$
$$= .21 kg/m^2$$

$$I_{cmst} = .21\ kg/m^2 - [5.18kg(.17m^2)]$$
$$= .060\ kg/m^2$$

$$Mass = \sum_{j=1}^{N} m_i$$
$$= 4.25kg$$

$$COM = \sum_{j=1}^{N} m_i d_i / m_{st}$$
$$= \frac{.525m/kg}{4.25\ kg}$$
$$= .124\ m$$

$$I_{ccst} = {}_{1/12} m_i(r_i^2 + h_i)$$
$$= 1/12(2.34)[(.074)^2 + (.125)^2]$$
$$= .0041\ kg/m^2$$

$$I_{fcs} = I_{ccst} + d_i^2(m_i)$$
$$= .0041\ kg/m^2 + .0625^2(2.34kg)$$
$$= .0132\ kg/m^2$$

$$I_{tfc} = \sum I_{fcst}(CC)$$
$$= .093\ kg/m^2$$

$$I_{cmst} = .093\ kg/m^2 - [4.25kg(.124m^2)]$$
$$= .028\ kg/m^2$$

Finalized Segments

For TF subjects, residual limb and, socket mass, center of mass and moment of inertia were combined to complete a final "thigh" segment. The characteristics of the componentry below the socket made up the "shank-foot" segment.

For TT subjects, the mass, center of mass and moment of inertia for the residual limb, socket and componentry below the socket were all combined to determine a finalized "shank-foot" segment.

Appendix II

Nomenclature

APPENDIX II

Nomenclature

C	- circumference
COM	- center of mass
CC	- circular cylinder
CP	- circular parabaloid
D	- density
di	- distance from the joint center to the COM of the ith cylindrical section
I	- moment of inertia
Iccst	- moment of inertia of the ith CC of the stump
Icmsk	- moment of inertia of the socket about the COM
Icmst	- moment of inertia of the stump about the COM
Icpst	- moment of inertia of the CP of the stump
Ifcst	- moment of inertia of the stump about the femoral condyle
Igtst	- moment of inertia of the stump about the greater trochanter
Isfcm	- moment of inertia of the shank-foot about the COM
Itfc	- total moment of inertia about the femoral condyle
Itgt	- total moment of inertia about the greater trochanter
kg	- kilograms
l	- liters
M	- mass
m	- meters
mi	- mass of the ith circular section of the stump
mst	- stump mass
sk	- socket
sf	- shank-foot
TF	- Transfemoral
TT	- Transtibial
V	- volume
X	- distance to the COM