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An Overview of Transfemoral Socket Designs

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AN OVERVIEW OF TRANSFEMORAL SOCKET DESIGNS

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This Independent Study, submitted by Bradley Wehe in partial fulfillment of the requirements for the Degree of Master of Physical Therapy from the University of North Dakota, has been read by the Chairperson of Physical Therapy under whom the work has been done is hereby approved.

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TABLE OF CONTENTS

Page

ACKNOWLEDGMENT v

ABSTRACT vi

CHAPTER

I. INTRODUCTION 1

II. LITERATURE REVIEW 4

History 4

Above Knee Biomechanics 6

Socket Designs 8

Suspension Systems 13

III. SUMMARY/CONCLUSIONS 15

REFERENCES 18

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ABSTRACT

In the field of amputee rehabilitation, there are many variables that determine if the outcome is one of success. A large part of the process evolves from the type of prosthesis that is recommended for the amputee. With the advent of new technology, we are faced with the problem of deciding on the proper components that will make up the prosthesis. The complexity of artificial limbs has increased over the past years as the concept of suction componentry has evolved.

We are now faced with multiple types and styles of transfemoral socket designs and suspensions. A literature review was performed utilizing the concepts of ischial containment sockets and the quadrilateral socket designs. Also discussed are the history, biomechanics, and suspension systems of each.

The result of this paper will be a better understanding of the transfemoral socket designs available for the amputee. This will assist the physical therapist or other health care providers in making the right decision for their patients.

CHAPTER I

INTRODUCTION

"Can it be saved" is the first inevitable question that a patient will ask a physician when presenting with a serious injury or disease of a limb. A thorough evaluation will follow along with consultations from varied specialists to assure the patient and family that every attempt has been made to avoid the amputation. Consultation may be sought from a vascular surgeon when major vessels are involved, or a diabetologist when infection is occurring in a diabetic. An infectious disease specialist has much to offer both pre-operatively and post-operatively. When tumors are involved, a surgical oncologist is suggested.[1]

In years past, there were often no alternatives to amputation. Through the progression of medical science, there are newer improved methods of fracture fixation, vessel and nerve repair, and limb reattachment. These techniques have provided many opportunities for limb salvage in cases destined for amputation. Because of the finality of amputation, both from a physical and psychological sense, these advanced procedures help to increase the patient's expectations of his/her outcome. The fact remains that prosthetic replacement following amputation falls short in the area of substitution for sensory and motor function. Limb salvage will always be the main goal, provided that it restores function better than the prosthetic replacement. There is, however, a significant patient population where limb salvage proves to be a hindrance and actually creates deficiency in maximizing functional ability.

Early amputation and prosthetic fitting are at times the preferred alternative to salvaging a questionably functional lower limb. A well-planned and executed amputation can remove a painful, dysfunctional limb, and allow rehabilitation with a prosthetic limb to a functional, painless state. Amputation surgery may be considered reconstructive surgery, with results similar to amputation of an arthritic femoral head and prosthetic replacement, such as a total hip arthroplasty.[1]

The evolution of new technology has drastically changed the field of amputee rehabilitation since World War II. Using the basic principles of support and stability for the above knee amputee, the progression of socket fit has changed from the plug fit to the new ischial containment sockets.[2] Leaders in the field, such as Sabolich and Long,[1] have been instrumental in the advancement of socket design. The plug fit design was initially used until the advent of suction sockets after World War II. In the early 1960s, the development of the total contact quadrilateral socket was established with or without suction suspension and was the socket of choice.[3]

At present, the ischial containment socket is the "state of the art" prosthetic design, with credit to Sabolich's Contoured Adducted Trochanteric-Controlled Alignment Method (CAT-CAM) Socket,[1] Ivan Longs' Normal Shape-

Normal Alignment (NSNA) Prosthesis,[1] and the more recent Narrow Medial-Lateral (Narrow M-L)[1] theory with a bony lock system.

This paper is a thorough literature review of the history, biomechanics, evolution of socket design, and suspension systems used today in above knee amputations.

CHAPTER II

LITERATURE REVIEW

History

Artificial limbs of different types, such as a forked stick, have been used since the beginning of mankind.[1] The earliest recorded use of a limb prosthesis is that of a Persian soldier, Hegesistratus, who Herodutus reported escaped about four hundred eight-four B.C. from stocks by cutting off his foot and replacing it with a wooden one.[1] The oldest known artificial limb in existence was a copper and wood leg unearthed at Capri, Italy, in 1858, which was supposedly made about three thousand B.C.[1] In 1529, Ambroise Pare, the father of amputation and a French military soldier, introduced the use of linen thread and ligatures as an alternative to boiling oil to cauterize bleeding. Morel introduced the tourniquet in 1674,[1] which allowed work in a relatively bloodless field with resultant rise in the survival rate. Sir James Syme reported the lower extremity Syme procedure for amputation at the ankle in 1843.[1]

The most common approach to the design of the transfemoral socket was the carved "plug fit" wooden socket with a conical interior shape. The weight of the amputee during the stance phase of walking and during standing was transferred to the skeletal system through the muscles and soft tissue about the thigh.[2] The transfemoral socket design[2] introduced by the University of California, Berkeley, about 1950 was shaped to permit use of the remaining musculature. Its design includes well-defined walls, and is known as the quadrilateral socket. The posterior wall was shaped to provide ischial-gluteal weight bearing along with leaving an air space between the distal end of the residual limb and the bottom of the socket. An air valve was installed in the medial wall. Due to excessive edema and dermatologic problems, further study led to the ischial containment socket design.[3] Thus, the second generation of transfemoral sockets was introduced.

In the early 1980s, Long, Sabolich, and others introduced designs known as NS/NA (normal shape-normal alignment), CAT-CAM (contoured, adducted trochanter-controlled alignment method), and Narrow ML (medial-lateral).[2] The one common feature of the above concepts was that the support of the amputee's body relied less on the ischial seat than the original quadrilateral design. The "ischial containment" sockets are now used in many areas of amputee rehabilitation, but additional research is needed to further document their efficiency and design.[3]

Immediately after World War II, wood and leather made up the majority of lower limb prostheses. These substances were found to be less than perfect. Carving and shaping of the older wood prosthesis, along with poor hygiene components of leather, led to Northrop Aviation's introduction of thermosetting resins for laminating tubular stockinette over plastic replicas of the residual limb to form components.[4]

The plastic laminating technique made total contact sockets practical, and was highlighted in 1972 when Snelson and Morney developed a method for vacuum-forming polycarbonate over a positive model of the limb.[2] This technique proved useful for check sockets and as an aid to teaching due to its transparent design.[5] Vacuum-forming polypropylene, which made this system acceptable for definitive use, was introduced in 1975 by Moss Rehabilitation Hospital in Philadelphia.[6]

Above Knee Biomechanics

The three major goals to strive for in prosthetic fitting are 1) amputee comfort when wearing the prosthesis, 2) efficient function with a minimal energy expenditure, and 3) acceptable cosmesis of the amputee's gait pattern and the prosthesis. The utilization of correct biomechanical technique is essential to control and direct the force vector present in an artificial limb. In the frontal plane, mediolateral stability of the pelvis is needed during midstance on the prosthetic side, along with conserving energy by minimizing lateral displacement of the amputee's center of gravity during gait. Anteroposterior stability is essential in the prosthetic knee between heel strike and heel off to allow the amputee to take a normal step forward with the non-amputated limb in the sagittal plane. These are the main biomechanical objectives in above knee prosthetics.[7]

Providing mediolateral stability of the pelvic during midstance on the prosthetic side is accomplished by a series of events. Static alignment is determined by positioning the prosthetic foot so that a plumb line dropped from the ischial seat of the socket will bisect the heel of the shoe.[8] This produces a varus moment at the ischium creating the pelvis to tilt towards the unsupported side.[9] The hip abductors fire to balance the force moment and displaced the femur laterally.

The lateral wall of the above knee prothesis is designed higher than the medial wall so it can resist the force exerted by lateral femoral displacement. The lateral wall of the socket needs to be adducted to re-establish the normal angle of the femoral shaft, thus placing the hip abductor muscles under tension. The lateral wall allows for even distribution of force over the lateral aspect of the residual limb.[8]

Socket relief is provided for the lateral distal end of the femur, with emphasis of force distributed between the greater trochanter and lateral distal end of the limb.[11] The medial socket wall must be high enough to apply counterpressure to maintain good contact between the femur and the lateral wall. The ischial containment socket give an increased medial force which utilizes the biomechanical advantage furnished by the design.[7]

The narrower the width of the walking base, the less the horizontal displacement of the body's center of gravity. By insetting the foot, the varus moment will increase thus decreasing energy consumption. But this also negatively impacts gait stability through a decreased walking base. Stability is enhanced through increasing the walking base and decreasing the varus moment by outsetting the foot.[10]

The sagittal plane forces provide anteroposterior stability of the prosthetic knee joint between heel strike and heel off.[8] This allows the amputee to take a normal step forward with the non-amputated limb. The ground reaction force must remain anterior to the prosthetic knee joint from heel strike to heel off on the prosthetic side.[9]

The trochanter-knee-ankle reference line (TKA) is utilized to achieve knee stability.[10] A line originating at the trochanter mark on the socket passes through the knee joint center and through the ankle joint center to allow the ground reaction force to remain anterior to the knee between heel strike and heel off on the prosthetic side. If the heel cushion, or plantar flexion bumper, is too stiff or there is an inadequate resistance to dorsiflexion, the ground reaction will tend to pass behind the knee center creating a flexion moment about the knee. This will result in a decrease of stability during the stance phase.

Socket Designs

The two basic functions of an above-knee prosthetic socket are support and stability.[11] The socket must be able to comfortably support the body while forces greater than the amputee's body weight are subjected to it. To stabilize the femur, the socket needs to be designed and aligned properly to control the hip and supporting musculature while allowing it to respond rapidly to muscle action.

There are five ideal concepts for the above knee design.[8] Force distribution is important and needs to be dispersed over the greatest area possible as pressure equals force divided by area. Pressure needs to be relieved over the neurovascular areas, but also application over specific areas, such as the femoral triangle, are imperative. When stabilizing forces are applied with a quadrilateral socket, application of force at the femoral triangle will keep the ischial tuberosity on the ischial seat.[11] Functioning muscles need to be on stretch to increase the mechanical advantage. Emphasis is placed on gluteus maximus and hip abductors. Lastly, proper contouring for functioning muscles is critical, especially when designing the proximal brim of the socket.

Until post World War II days, the old plug fit socket design was utilized for the majority of amputees.[1] Plug fit designs lacked any special contouring or relief areas for musculature. It was cylindrical in shape and the design resembled a "cork in a bottle" type of suspension. These are rarely seen now as the technology and force vector studies have proven them inefficient from an energy consumption and comfort standpoint.

There are many socket designs available for the above-knee amputee, but the quadrilateral design has been proven to provide the most support and stability.[1] The term quadrilateral refers to the appearance of the socket when viewed in the transverse plane. There are four distinguishable sides or walls of the socket. Weight bearing is achieved primarily though the ischium and the gluteal musculature. The lateral wall is designed to support the femoral shaft, and allows the hip abductor muscles to contract effectively. Medial-lateral stabilization occurs on the lateral distal wall and the medial proximal wall.[12] The medial wall is designed to compress the limb against the lateral wall and to contain medial thigh soft tissue.

The posterior wall provides the major area for weight bearing on the ischial tuberosity and gluteus maximus muscles along with supporting the femoral shaft so that hip extensors can contract efficiently. The anterior wall functions to hold the residual limb back on the ischial seat, thus acception of the majority of body weight at stance phase.

The quadrilateral design has a narrow anterior-posterior dimension and the anterior wall is higher than the posterior wall. Overall, the socket has initial flexion to improve the ability of the amputee to control knee stability at heel contact and to help in minimizing the development of lumbar lordosis at toe off.[13] Adduction is built into the design to enhance the efficiency of the hip abductors.

Proper socket contours for actively functioning muscles, such as the rectus femoris and gluteus maximus, will affect tracking of the prosthesis during swing phase. If the anterior-posterior dimension of the socket is too tight, then muscle activity in the swing phase of gait can lead to undesirable socket rotation which will appear clinically as swing phase whips.

The concept of the United States quadrilateral socket was borrowed from Europe and refined through biomechanical analysis and research conducted here.[12] In the 1980s, European style quadrilateral brims were available in the United States. These featured a design that allowed socket walls to be smoother and less abrupt.[12] The medio-proximal wall was lowered which increased perineum comfort. A larger anterior-posterior dimension was noted, along with a narrower medial-lateral view. The biomechanical principles remained the same, but these changes in socket design led to the ischial containment socket theory and initiated new concepts in transfemoral socket theory.

The quadrilateral above-knee socket design has been the mainstay of prosthetic care for an individual with an above knee amputation since 1949, when Inman and Eberhart brought the design to the United States.[13] Careful analysis of support, stability, and comfort theories indicate that a change in socket configuration may be necessary. Lehneis[12] stated that ischial weight bearing on the horizontal posterior wall of the above knee quadrilateral socket is ineffective in all but the midstance phase of gait due to the motion of the femur and socket in relation to the hip joint. He also questioned the necessity of ischial weight bearing in total contact sockets, where Pascal's Law applies in distributing forces equally around a fluid mass.[13]

Long[13] challenges the concept of ischial weight bearing from the point of view of its inability to stabilize the femur when the gluteus maximus fires. When the femur exerts force against the lateral wall in weight bearing, the quadrilateral socket moves laterally immediately, because the ischium has no effect of stopping this shift.[13]

Sabolich[12] is critical of the drift of the ischial tuberosity on the posterior wall of the quadrilateral socket with resultant lateral trunk leaning by the patient in a effort to stabilize the pelvis. He also comments that a tangential force, at best, is possible with the curved ischial tuberosity resting on the neurovascular bundle anteriorly, with the purpose of pushing the ischial tuberosity up onto the ischial seat due to concerns for the circulatory status of the residual limb.[14]

Design characteristics of the ischial containment socket include a narrow medial-lateral dimension, containment of the ischial tuberosity with a portion of the ramus of the ischium in the socket, slanting of the posterior wall, a higher lateral wall with medially directed forces proximally and distal to the greater trochanter to create a locking or wedging effect within the socket to stabilize the femur, and adduction of the femur.[14] The narrow medial-lateral design produces increased stability through bearing pressure against the skeletal elements, thus reducing motion lost through soft tissue. Muscular function is not inhibited by the crowding effect of a narrowed anterior-posterior dimension.

Containment of the ischial tuberosity and a portion of the ramus prevents lateral shifting of the socket and increases comfort for the amputee.[3] The posterior wall is slanted forward and downward to increase comfort at heel strike and heel off. The high lateral wall with medially directed forces borne by the ischial tuberosity, creates a three-point pressure system to lock the femur into adduction and reduce motion that can occur when the ischium is free to shift. The ischium, trochanter, and latero-distal aspect of the femur provides a much more stable mechanism for acceptance of perineal biomechanical forces, thus resulting in the bony-lock system.[4]

Regardless of the fitting method used, the socket for any amputee must provide the same overall functional characteristics, including comfortable weight bearing, stability in the stance phase of gait, and a narrow based gait pattern. In addition, it must also furnish as normal a swing phase as possible consistent with function.[11]

Suspension

Prior to designs based on biomechanical principles, suspension of lower limb prosthesis presented problems. Until the introduction of the pelvic band around World War II, over-the-shoulder suspenders were used universally.[14] These are rarely seen as a means of suspension today. Hip joint, pelvic band, and waist belts are used to increase the medial-lateral stability when the hip abductors are weak.[5] Geriatric amputees are the most common users of hip joint and pelvic band suspension.[1] Short limb amputees also are indicated for this type of suspension.

Silesian belts are the second most common use of suspension in the 1990s.[14] These allow for extra suspension commonly used to supplement suction suspension. They also give additional rotational stability, but do not assist in medial-lateral stability. The belt attaches at a pivot point at the lateral aspect of the socket, and extends around the back over the iliac crest to the anterior midline.

The most common type of suspension for transfemoral amputation is suction socket suspension.[14] Suction suspension refers to the technique of maintaining the prosthesis by negative air pressure in the socket itself, which holds the prosthesis on during swing phase. This is accomplished by the use of an air valve at the distal end of the socket and a well contoured socket that fits directly around the amputee's skin of the stump to form a seal. No prosthesis liner socks are used with full suction suspension as air would leak around the sock.

Suction suspension eliminates pistoning of the residual limb, improves proprioceptive input, allows for a total contact fit, and is lighter.[5] It is more difficult to don and doff, however. Partial suction suspension may also be utilized, in which the amputee uses the suction liner with a prosthetic sock along with a Silesian belt.

Suction sockets are indicated for average to long above knee amputees who have stable residual limbs.[14] Although the term "suction" is employed to describe this type of suspension, the intimate socket fit around the musculature of the residual limb is the most important factor in socket design. Suction will not suspend a prosthesis properly if the amputee's muscles are not properly accommodated in the socket.

CHAPTER III

SUMMARY/CONCLUSIONS

Through the evolution of new technology, we have observed the field of amputee rehabilitation grow to new heights. Amputees are experiencing a higher level of independence and comfort than ever before. Technology and advanced biomechanical studies have continued to take us beyond the basic plug fit prosthetic design that was once state-of-the-art and led us into the containment sockets. More emphasis is put on socket design and suspension to achieve additional comfort and performance with less energy expended.

The two basic functions of an above knee prosthetic socket are support and stability. There are multiple techniques that can be utilized to achieve these ultimate goals. Force distribution is important and needs to be dispersed over the greatest area possible. Pressure must be relieved over neurovascular areas, but also must be applied over specific points for stability. Functioning muscles need to be put on anatomical stretch to enhance mechanical advantage. These are all examples of ideal concepts for the above knee design.

The three main types of transfemoral socket designs that were discussed are the plug fit, quadrilateral and ischial containment sockets. The quadrilateral socket design utilizes an ischial weight bearing concept and does not have a bony lock system. The posterior wall provides for the majority of the weight bearing on the gluteus maximus and ischial tuberosity. Lehneis[12] stated that independent weight bearing is in effect in all phases of gait except at midstance. This is due to the motion of the femur and socket in relation to the hip joint. Long[13] also challenges the concept of ischial weight bearing as femur stabilization is poor when the gluteus maximus fires.

Ischial containment socket designs utilize a narrow mediolateral dimension along with containing the ischial tuberosity and a portion of the ramus of the ischium inside the socket. By slanting the posterior wall and having a higher lateral wall, a locking or wedging effect is achieved to stabilize the femur. The narrow ML design produces increased stability by bearing pressure against the skeletal elements, thus reducing motion lost through soft tissue. The three-point pressure system used to lock the femur into adduction is used to reduce motion and increase stability.

Regardless of the fitting method used, the socket for any amputee must provide the overall same functions. There are multiple techniques and styles that may be used when constructing an artificial limb. Despite the varied systems utilized, there are three major goals for which to strive. These are: 1) amputee comfort, 2) efficient function, and 3) acceptable cosmesis. The main biomechanical objectives need to be applied regardless of which method is used. Proper placement of ground reaction force vectors along with stabilization of the pelvis is essential in meeting the goals.

In the years past, there were no alternatives to amputations. With the advances in medical science and technology, the field of amputee management has grown. The prosthetic replacements that were once state-of-the-art are now deemed archaic. Through these advances, we are able to have a variety of choices when making our decision on the appropriate socket design for the amputee. "Can it be saved" certainly may be the first question that a patient may ask when presenting with a serious injury but, hopefully, when the decision is "no," the previously mentioned techniques will be applied to assure the amputee ultimate function and enhanced overall independence.

REFERENCES

1. Bowker J, Michael J, eds. Atlas of Limb Prosthesis. St. Louis, Mo: Mosby Year Book; 1992.
2. Prosthetic-Orthotic Center. Lower Limb Prosthetics for Therapists. Northwestern Medical School; 1991.
3. Berry D. Non ischial weight bearing soft socket designs. Clin Prosth Orth. 1986;2:43-48.
4. Prosthetics Consultants. Narrow M-L Socket Design. New York University; 1991.
5. Murphy E. Sockets, linings, and interfaces. Clin Prosth Orth. 1990;3:4-6.
6. Horowitz J, Leflan M. Alternative casting techniques for narrow M-L sockets. Clin Prosth Orth. 1988;4:66-70.
7. Pritham C. Biomechanics and shape of the above-knee socket. Clin Prosth Orth. 1990;14:9-21.
8. Gage J, Hicks R. Gait analysis in prosthetics. Clin Prosth Orth. 1990;3:17-21.
9. Nissan M. Initiation of gait in lower limb amputees. J of Rehab. 1991;2:1-12.

10. Muilenberg A. Basic changes in lower limb prosthetics. Phys Ther. 1985;9:5-6.

11. Radcliffe. Fitting AK prosthesis. Clin Prosth Orth. 1985;2:35-60.

12. Lehneis R. Beyond the quadrilateral. Clin Prosth Orth. 1985;9:6-9.

13. Long I. Normal shape-normal alignment prosthetics. Phys Ther. 1983;9:9-14.

14. Anderson M, Bray J, Hennessy C, Sollars R. Prosthetic Principles--Above Knee Amputations. Springfield, Ill: Charles C Thomas Co; 1960.