Prosthetics and Orthotics International, 1993, 17, 21–26

Movement of the tibial end in a PTB prosthesis socket: a sagittal X-ray study of the PTB prosthesis

M. LILJA, T. JOHANSSON and T. OBERG

Department of Biomechanics and Orthopaedic Technology, University College of Health Sciences, Jönköping, Sweden

Abstract

To investigate the movement of the tibial end in the sagittal plane in the PTB prosthetic socket during a gait cycle, 7 patients with a median age of 72 years were examined using X-ray technique. The gait cycle was reduced to four different static positions: heel contact, mid-stance, push-off and swing phase. The mean value of tibial movement in the socket in the anteroposterior direction was 2.2 cm, in proximodistal direction 2.8 cm, and the total sagittal movement during the whole gait cycle was 7.5 cm. The results indicate that one factor affecting the magnitude of the movement was the prestretching of soft tissues. All the patients who experienced a good prosthetic fitting had their soft tissues prestretched. The extreme dorsal and proximal positions of the tibial end during the gait cycle was in the swing phase position. The extreme distal position occurred somewhere between mid-stance and push-off. The extreme anterior position of the tibial end was seen during heel contact. This study has shown the magnitude of the movements in a PTB socket during a simulated gait cycle. The study has given hints on factors affecting prosthetic fitting, and further research within this field might provide indications of how to optimise socket shape to give maximal patient comfort.

Introduction

Pain in the amputation stump is a major problem for both the patient and the prosthetist (Mattson, 1980; Whylie, 1991; Troup, 1988). According to Persson and Liedberg (1983) 20% of all amputees have stump pain. Other frequent problems are secondary wound healing and adherent scars (Yaramenko and Andruhova, 1986; Persson and Liedberg, 1983). Yaramenko and Andruhova (1986) found a frequency of 11.5% and Persson and Liedberg (1983) 9% of secondary wound healing in trans-tibial amptees.

When the soft tissues adhere to the skeleton, movement between the tibial end and the prosthetic socket can be a factor causing problems. The PTB socket and the PTB suction socket allow movement of the tibial end during walking. For PTB prostheses Eriksson and Lempberg (1969) showed a movement in the proximodistal direction of 2.25 cm. For PTB suction prostheses the movement in the proximodistal direction was found to be 1.1 cm (Grevsten and Eriksson, 1975). In the authors' opinion there is a connection between pain, adherent scar and the movement of the tibial end in the socket. They also believe that the skeletal movements can be one explanation of stump pain. When the soft tissues adhere to the bone, bone movement will stretch the skin. This stretching can result in laceration and in the worst case, necrosis (Lilja and Johansson, 1992).

It is even more important to decrease the movements in the socket, as it is known that skin damage increases with time in patients with prostheses (Persson and Liedberg, 1983).

The aim of this study was to examine the tibial movements in a PTB prosthesis. In contrast to earlier studies, the simulation of the different phases of the gait cycle in the present

All correspondence to be addressed to Magnus Lilja, B.Sc. C.P.O., Department of Biomechanics and Orthopaedic Technology, University College of Health Sciences, P.O. Box 1038, S-551 11 Jönköping, Sweden.

study is more representative of all the tibial movement during a normal gait cycle with a PTB prosthesis. The investigation was limited to trans-tibial prostheses. In studies of tibial movement in PTB prostheses, X-ray is a well documented method (Newton *et al.*, 1988; Baumgartner *et al.*, 1980; Haslan *et al.*, 1983; Bao-Shan *et al.*, 1977).

Material and method

Seven unilateral trans-tibial amputees with a median age of 72 (61–79) years, were examined. All patients, 2 women and 5 men, had PTB prostheses. The main diagnoses were diabetes mellitus in 5 and arteriosclerosis in 2 of the cases. All amputations were performed with a long posterior flap. Three of the patients had used their prosthesis for less than 6 months and none had used it for more than 1 year. The patients were consecutively selected from a population of trans-tibially amputated patients at the orthopaedic clinic of the county hospital of Jönköping, Sweden. The averge stump length was 14.1 (10–20) cm. The average width over the condyles was 9.5 (9–12) cm. Four out of seven patients perceived a good prosthetic fit. Three of these four patients had the soft tissues prestretched. This prestretching was carried out by the patient himself when putting on the prosthesis. The three patients who did not prestretch the soft tissues perceived the prosthetic fitting as poor.

Experimental setup

The patient was examined in four different positions representing four phases in the gait cycle: heel contact, mid-stance push-off and finally swing phase. To simulate heel contact and push-off the floor was tilted at an angle of 15 degrees. The patient could stand vertically on either the toes or the heel on the tilted floor and then load with the whole body weight. The floor reaction force affected the knee joint with an extending or flexing moment exactly as in push-off and heel contact respectively during normal gait. In mid-stance the patient was standing on a plane floor.

Normally the floor reaction force is about 75% of the body weight in mid-stance, and about 120% of the body weight in push-off and heel contact (Cunningham, 1958), but in the authors' simulation the floor reaction force was equal to the body weight in all these positions. To simulate the swing phase the prosthesis was positioned at an angle of 45 degrees relative to the floor.

X-ray examination

All X-ray pictures were taken in a sagittal projection. To determine the scale a metal measuring device was used. The roentgenological examination was performed according to a standard procedure. To obtain high contrast radiographs, all prostheses were prepared with a thin steel wire, running along the inner surface of the external prosthetic socket. This procedure was undertaken to simplify the measurement of the tibial end positions.

Calculation of movements

To calculate the total movement of the tibial end, the four positions of the tibial end were marked in a co-ordinate system. The four positions were joined by straight lines, forming a polygon. The perimeter of this polygon is almost equivalent to the total movement in the sagittal plane during the gait cycle. This perimeter describes the minimum movement that the tibia can perform in the sagittal plane.

Results

Overview

In all cases a distinct and uniform motion of the tibia in the prosthetic socket was observed during simulated gait. All graphs of the tibial movement showed similar patterns of movement. During the swing phase large movements of the tibial end in a posterior and proximal direction were observed. Then the movement changed into an anterior and distal direction. The pattern of movements and the magnitude of the movements are shown in Figure 1.

Total movement in the sagittal plane

The average total movement in the sagittal plane was 7.5 (5.3-9.6) cm. The movements in the individual patients are shown in Figure 2. The largest movement occurred in the anterodistal direction between push-off and swing phase. This movement was seen in all but one patient, who showed the largest movement between swing phase and heel contact.

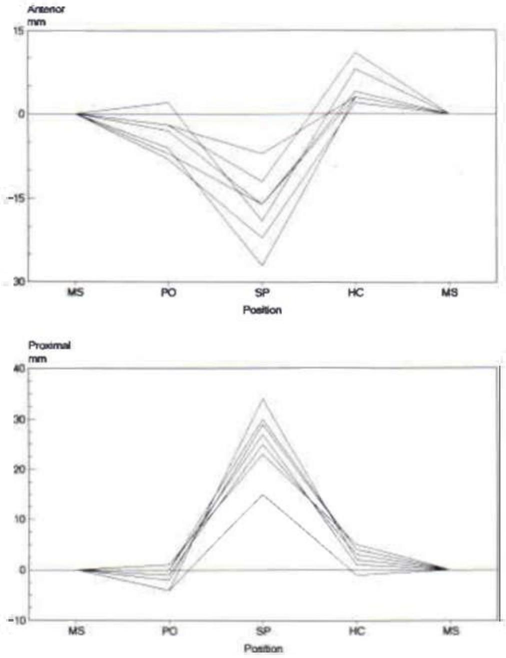

Figure 1. The pattern of movements and the magnitude of the movements of the tibial end in a PTB prosthesis socket in seven patients. HC=heel contact, MS=mid-stance, PO=push-off, SP=swing phase.
Top: Movements in the anteroposterior direction.
Bottom: Movements in the proximodistal direction.

Movement in anteroposterior direction

The average total horizontal movement of the tibial end was 2.2 (1.0-3.0) cm. The most posterior position of the tibia was seen during the swing phase and the most anterior position at heel contact. The largest movements of the tibia in the prosthetic socket occurred when the prosthesis was lifted up and swung forward to heel contact. The cumulative movement of tibia in anteroposterior direction during a whole gait cycle was 4.5 (2.1-6.0) cm.

Movement in proximodistal direction

The movement of the tibial end in the proximodistal direction was larger than the movements in the anteroposterior direction. The mean value was 2.8 (2.0-4.0) cm. The extreme proximal and distal positions vary between swing phase and push-off or swing phase and mid-stance. The mean value of the cumulative total movement in the proximodistal direction during a whole gait cycle was 5.7 (4.2-8.1) cm.

Discussion

Methodological considerations

The four positions were chosen for the following reasons: at heel contact and push-off the floor reaction forces are maximal at the same time as the lever arms to the knee are as long as possible and consequently the moments of force about the knee axis at a maximum. Mid-stance was chosen as a neutral position when the floor reaction force passes through the knee joint. The swing phase finally was chosen because it is the position when the weight of the prosthesis has the maximum effect on the stump at the same time as the triceps surae muscle is relaxed (Grevsten and Ståhlberg, 1975). It was attempted to simulate the direction of the floor reaction forces, especially in relation to the knee joint. The dynamic component of the floor reaction force could not be simulated in the static model. The heel contact and push-off phases in normal gait give a dynamic contribution to the floor reaction force which can reach about 120% of the body weight. Normally, at mid-stance, the floor reaction force is about 75% of the body weight (Cunningham, 1958; Lamoreux, 1985). In all positions the assumption was made that the floor reaction force was 100% of body weight. In spite of that, the model is regarded as acceptable as an approximation of slow gait with fairly moderate dynamic contributions to the floor reaction forces. Furthermore, the patient himself decided how much pressure he could tolerate by leaning the body backwards or forwards. The load level in this study was as close as possible to the tolerance limit of the patient, which means that the patients compensated for load differences.

Total movements

The movements of the tibia in a prosthetic socket have been studied in a few earlier studies (Eriksson and Lempberg, 1969; Grevsten and Eriksson, 1975). In this study the pattern of the tibial movement was uniform for all the subjects during the gait cycle. This motion pattern is regarded as valid for the majority of PTB prostheses with a soft socket. However, the magnitude of movement differed from one patient to another. Prestretching of the soft tissues seemed to be a factor affecting the magnitude of movement. In all cases except one, the largest movement was observed

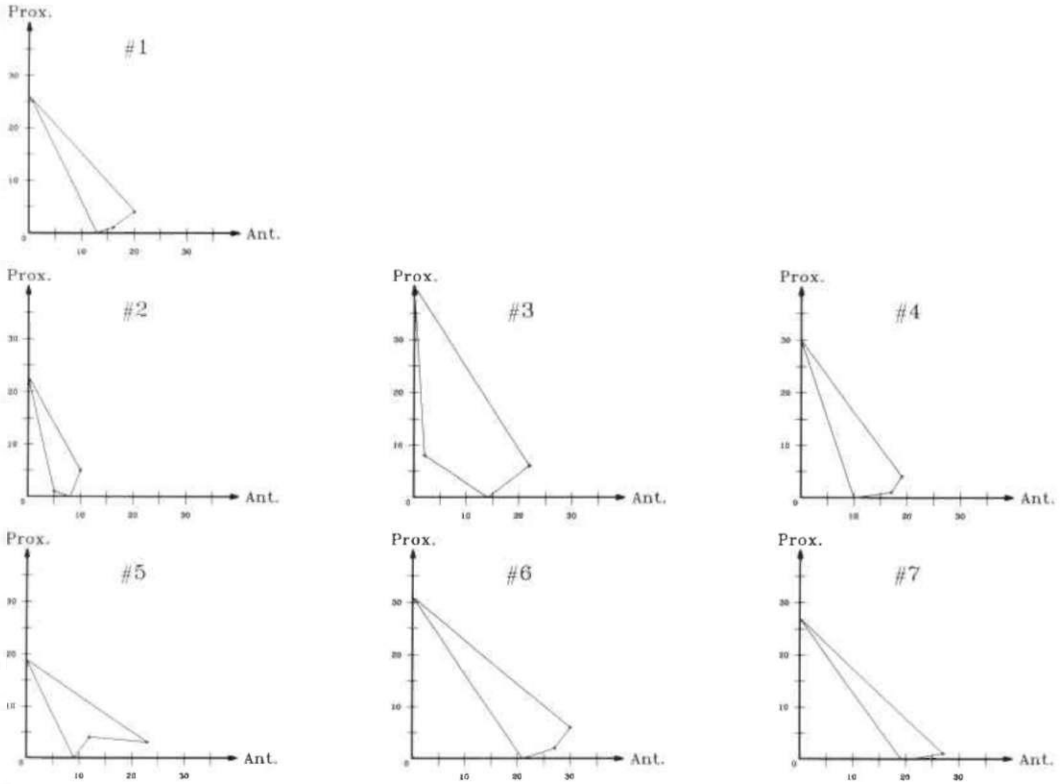

Figure 2. Movements of individual patients. # indicates case number. The movements in anteroposterior direction are indicated on the X-axis, those in proximodistal direction on the Y-axis.

between the swing phase and heel contact. In cases where the soft tissues were prestretched the total movement was reduced. Standing in the swing phase produces maximal moments of forces about the knee axis caused by the weight of the prosthesis. In this position the lever arm is as long as possible, the triceps surae muscle is relaxed (Grevsten and Ståhlberg, 1975), and the tibia is in a maximal dorsal and proximal position in the prosthetic socket. The most distal positioning of the tibial end occurs at some time between push-off and mid-stance. The most anterior position was, as expected, at heel contact.

Clinical implications

Two extreme cases can be theoretically considered: a) the prosthesis is anchored to the bone, and all energy and all movement is transferred to the prosthesis, and b) the movement of the tibia in the soft tissues is so large, that no energy and no movement is transferred. In reality there is a situation somewhere between these two extremes.

In the first case there is no energy loss. This is, of course, the ideal situation. In the second case all movement energy is lost in the soft tissues of the stump, i.e. the effort made by the patient will give no output as gait movements. This can be seen as the worst case of bad prosthetic fitting. The larger the movements of the bone and in the soft tissue, the larger the energy loss. The movement of the tibial end describes a polygonal figure (Fig. 2). The surface of this polygon will, in some respect, reflect the magnitude of the energy loss. The exact relationship between the movements of the tibial end and the energy loss cannot be easily described due to the non-linear viscoelastic properties of the soft tissues (Fung, 1987).

In an earlier study on PTB suction prostheses (Grevsten and Eriksson, 1975) the movements were twice as large in the proximodistal direction but similar in the anteroposterior direction as those found in this present study. It is believed that this can be explained by an inadequate suspension in the PTB suction prosthesis (Grevsten, 1978). The total amount of movement of the distal tibial end in the sagittal plane was as large as 7.5 cm. According to an earlier study of 22 patients provided with PTB prostheses (Öberg *et al.*, 1992) the median distance that the patients walked every day was 1,550 m. The stride-length was about 0.5 m. Thus the patients took about 3,100 steps per day. If one looks at the tibial movements inside the PTB socket it can be seen that the tibial end moves about 7.5 cm extra for each step. During one day the tibial end will move approximately 232.5 m extra inside the PTB prosthesis.

Lower limb amputees are generally old people with a mean age of 60–80 years in the vascular cases. They often have other diseases as well, and many of them have small residual capacity. Whatever the magnitude of the energy loss, it will influence the clinical outcome and performance of the patient. The body has a preference for the least energy consuming strategy, and energy loss may be the critical factor in deciding if the patient will be a walker or wheelchair case.

The movement of the tibial end in the soft tissues also greatly influences proprioception and kinesthesis. The position of the limb is mainly registered by proprioceptors in or nearby the joints. If the joint movement is not followed by a corresponding movement of the prosthesis, it is impossible to know the exact position of the foot, and consequently there will be an inaccurate gait pattern. In the ideal case all movement is transferred to the prosthesis, in the worst case no joint movement is transferred.

If numbers are considered the clinical results of the horizontal movement can be seen. One of the patients had a horizontal tibial end movement of 3.0 cm when he moves from swing phase to heel contact. The length of his lower limb was 42 cm from the knee down to the heel. The length of the amputated tibia was 14 cm. The difference of the actual position of the prosthetic foot and the position expected by proprioception and kinesthesis was 9 cm. Clinically, for every step the patient takes he has to produce a larger motion in the knee and the hip to compensate for the movements of the tibia inside the socket. This motion will lead to extra learning problems in prosthetic training although probably the patients will learn relatively quickly the new position of the prosthetic foot.

If the skin sticks to the walls of the prosthetic socket, all movement takes place in the soft tissue, and no friction energy is released in the contact between the skin and socket. If however there are adherences between the tibial end, the soft tissues and the skin, the movement of the tibial end are transferred directly to the skin, with friction and danger of friction wounds (Lilja and Johansson, 1992). Sometimes it can be observed that the outer skin layer of the stump is removed. This could be an effect of the friction between the stump and the socket. Thus even in this respect, movements of the tibial end in the socket can be clinically important.

At present knowledge concerning soft tissue mechanics of the amputation stump is scarce. The result of this study strongly indicates a need for further studies of stump and soft tissue mechanics to elucidate the relationship between bone movement, soft tissue deformation, energy loss and energy transferred to the prosthesis. It also indicates a need for studies where bone anchored prostheses are compared with conventional prostheses. Such studies can give hints on the magnitude of energy loss in the soft tissue.

Comparisons of tibial motion and socket types would result in further understanding of this problem. The results of such studies might lead to recommendations of what kind of prosthetic socket should be used when problems occur, (for instance adherent scars, oedema, hyperaesthesia, different diseases e.g. vascular or dermal diseases etc.) or to present problems.

Studies of tibial movement in prosthetic socket might explain some problems of pain in patients with adherent scars, and indicate how to optimize the form of the socket to give maximal comfort to the patient.

From the theoretical considerations above, it can be concluded that movements of the bony end in amputation stumps may have important clinical implications, and may also be a measure of good or bad prosthetic fitting.

REFERENCES

Bao-Shan J, Villanueva R, Dodd GD (1977). A new radiological technique in evaluation of prosthetic fitting. *Radiology* **122,** 534–535.

Baumgartner R, Langlotz M (1980). Amputee stump radiology. *Prosthet Orthot Int* **4**, 97–100.

Cunningham DM (1958). Components of floor reactions during walking. Berkeley: University of California. Prosthetic Devices Research Project 14.

Erikson U, Lempberg R (1969). Roentgenological study of movements of the amputation stump within the prosthesis socket in below knee amputees fitted with a PTB prosthesis. *Acta Orthop Scand* **40**, 520–529.

Fung YC (1987). Mechanics of soft tissues. In: handbook of bioengineering./edited by Skalak R, Chien S. – New York: McGraw-Hill. 1.1–1.12.

Grevsten S (1978). Ideas on the suspension of the below-knee prosthesis. *Prosthet Orthot Int* **2**, 3–7.

Grevsten S, Erikson U (1975). A roentgenological study of the stump-socket contact and skeletal displacement in the PTB-Suction Prosthesis. *Upsala J Med Sci* **80**, 49–57.

Grevsten S, Ståhlberg E (1975). Electromyographic studies of muscular activity in the amputation stump during walking with PTB- and PTB-suction prosthesis. *Upsala J Med Sci* **80**, 103–112.

Haslam TP, Wilson MT, Urlaub MJ, Borrell RM (1983). The use of X-rays and xerograms in prosthetics. *Orthot Prosthet* **37(1)**, 32–35.

Lamoreux L (1985). Measurement and analysis of ground reaction forces. In: (Proceedings of the 1985 Uppsala Gait Analysis Meeting), Gait analysis in theory and practice./edited by Lanshammar H. – Uppsala University, Sweden: Uppsala University, Department of Technology, p47–69.

Lilja M, Johansson T (1992). Adherent cicatrix after below knee amputation (Accepted for publication), *J Prosthet Orthot.*

Mattson HS (1980). Fantombesvär och andra smärttillstånd. (Phantom pain and other pain conditions) In: Amputationskirurgi och proteser. (Amputation surgery and prostheses.) 2nd edition. edited by Hierton T. — Sweden, Falkoping: Gummessons tryckeri AB p 88-94. (In Swedish.)

Newton RL, Morgan D, Schreiber MH (1988). Radiological evaluation of prosthetic fit in below knee amputees. *Skeletal Radiol* **17**, 276–280.

Persson BM, Liedberg E (1983). A clinical standard of stump measurement and classification in lower limb amputees. *Prosthet Orthot Int* **7**, 17–24.

Troup IM (1988). Pre-operative and post-operative care: stump environment. In: Amputation surgery and lower limb prosthetics./edited by Murdoch G, Donovan R. — Oxford: Blackwell Scientific Publ. p 21–28.

Whylie B (1991). The psycho-social impact of amputation. In: Clinical aspects of lower extremity prosthetics./edited by Weber D. — Winnipeg: Elgan Enterprise. p 3–9.

Yaramenko D, Andruhova RV (1986). Below-knee amputations in patients with vascular disease and prosthetic fitting problems. *Prosthet Orthot Int* **10**, 125–128.

Öberg T, Lilja M, Johansson T, Karsznia A (1992). Klinisk utvärdering av proteshylsor. (Clinical evaluation of prosthetic sockets) Jonkoping, Sweden: University College of Health Sciences, Department of Biomechanics and Orthopaedic Technology. (In Swedish.)