PROSTHETIC-FEET AND SHOES : THEORY AND PRACTICE.

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INTRODUCTION

Nowadays we can chose a great number of prosthetic-feet in manufacturing leg prosthesis. The material and functional characteristics differ strongly. There are flexible respectively rigid prosthetic-feet, with and without ankle-functions.

Which prosthetic-foot belongs to which clinical starting-situation? On the basis of scientific literature and the information of the manufacturer of prosthetic-feet no clear answer is given on this question. In practice everybody, physician and prosthetist, acts for the greater part on his own insight. So one person prefers a Sach-foot in case of a geriatric patient with a below-knee amputation and the other a prosthetic-foot with ankle-axis. With respect to the use of shoes in combination with a prosthetic-foot concrete data are missing as well.

Which shoe belongs to which prosthetic-foot? Usually the patient decides the choice of the shoes. Only the heelheight (2- 3 cm.) of the shoes is often prescribed. Based on a fixed heelheight the prosthesis is aligned. Regularly however we meet users of a legprosthesis who wear shoes with different heelheights. Biomechanically there is a lack of abjective information about the combination of prosthetic foot/shoe. Mainly subjective considerations determine the way in which prosthetic-feet and shoes are used. To try to get a better insight in the field of the combination of prosthetic foot/shoe the Research department of the "Roessingh" Rehabilitation Centre and the tutorial Biomedical engineering of the Twente University started a pilot study.

QUESTIONS OF THE PILOT STUDY

We werw interested in which way 6 different types of prosthetic feet and 2 models of ready-made shoes during walking of 5 good walkers with a below-knee amputation manifest with respect to:

1. steptime parameters (time-, distance- and symmetry-factors).
2. dorsal/plantar flexion of the "ankle".
3. extension/flexion of the knee.
4. muscle activity of the upperleg musculature.

PATIENTS AND METHOD OF THE STUDY

Five patients, 22, 29, 31, 52 and 58 years old, were selected on the following criteria: below-knee amputation left; traumatic cause; shoe size 43; no stumpcomplaints; well fitted stumpsocket; good walking performances. The prosthetic-feet studied were the Otto Bock single axis-foot, the Rax- foot, the Otto Bock Dynamic foot, the Endolite foot, the Otto Bock Sach foot and the Seatle foot. We suspected that, besides the type of prosthetic-foot, the characteristics of the shoes also would influence the magnitudes to measure. Therefore our patients walked with the different prosthetic-feet in combination with as well a pair of flexible shoes as a pair of stiff leather shoes (heelheight 2 cm.). In practice both types of shoes are alternately used by the same person. During walking with a comfortable walkingspeed the next parameters are measured:

a. time-, distance- and symmetry factors during walking over a distance of 10 meters.
b. the average moving patterns of the knee of the affected side of the patient and the movement of the prosthetic foot at the level of the original ankle-axis. These averages have been measured during 10-20 walking cycles.
c. at the same time the EMG-muscle activation patterns of the m. Rectus femoris and the m. semitendinosus have been registrated .

The measurement took place in the gate-laboratory, with help of the Quark-system (Kleissen, 1989).

First the measurements were performed in all patients, wearing their own prosthetic-foot. After that, another prosthetic-foot was directly fixed at the prosthesis by a prosthetist. The patients went home for a week to learn using the new prosthetic-foot in their daily living. After this week the patient came to the laboratory again and the same measurements wer done. The next prosthetic foot was fixed (and so on). This was continued until the last prosthetic-foot had been investigated. After that the patient received his original prosthetic-foot again. Because of technical considerations the prosthetic-feet were not assigned randomly. All measurements have been using to two types of shoes.

RESULTS

Time, distance and symmetry factors

Walking speed: the average walking speed (concerning 10 meters) was not much different from normal and did not depend on the type of the prosthetic foot used. There was no influence from the type of shoes on the walking speed either.

Symmetry of the stance-times: all patients walked somewhat asymmetrically. The stance-phase of the healthy leg was little longer than the stance-phase of the prosthetic leg, exept in walking with a Sach-foot. Here we saw that in 3 of the 5 patients the stance-phase of the prosthetic leg was longer. There was no influence to show of the type of prosthetic foot respectively shoes on the asymmetry.

Movement of the "ankle" of the prosthetic foot

At heelstrike the prosthetic foot touch down on the heel and the prosthetic foot will make some kind of plantar flexion movement till the sole of the shoe is flat on the ground. The maximal plantar flexion angle, averaged over 5 patients,is presented in fig. 1 dependent on the type of shoes and prosthetic feet. One can see, that the rigid Sach-foot moves a few degrees, like the Seatle-foot. The Endolite-foot and the Otto Bock-single-axis-foot moves 4 up to 5 times more. Another thing you see is, that the leather shoe seems to activate systematically more plantar flexion. In the late stan-cephase just before toe-off, a maximum in dorsal flexion of the foot occurs as a result of the deformation of the prosthetic foot. The size of this dorsal flexion is mentioned in fig. 2. Again it is clear that the Sach- foot deforms only a little, but now the Seatle-foot is much more flexible than the Sach-foot. It remarks that now in all the prosthetic feet the leather shoes gave less dorsal flexion than with the sportshoes.

Movement of the knee-joint

A disadvantage in the method of measuring goniometersignals is, that strict repositioning of the goniometers is not possible.

Moreover a gaugers measurement, set up with the goniometers and the patient in a standard position, gives only an approximation of the relation between the joint angle and the angle of the goniometry, because a standard position is not exactly to define. On behalf of the processing of kneegoniometer- signals it is assumed, that the knee-angle at heelstrike in one patient is well reproducible. This value can be used as a reference.

This idea is derived from the fact that the moving patterns of normal persons are well reproducible in walking with a comfortable walking speed (Inman, 1981; Winter, 1987). It appears from the literature mentioned, that in normal persons the knee is nearly extended at heelstrike (the knee-angle shows less than 3 degrees flexion). In normal persons the maximal kneeflexion is about 20 degrees in the early stance-phase. The users of a prosthesis show a smaller maximal kneeflexion. Maximal this was averaged over all prosthetic feet by using leather shoes. This is presented in table 1. Moreover it appears, that kneeflexion in the stance-phase is significant bigger walking with leather shoes than walking with sportshoes (see table 1). From this it follows that kneeflexion in walking with sportshoes becomes still smaller with respect to the normal kneeflexion. As one can understand from table 2, the maximal kneeflexion is smaller in the stance-phase for all patients walking with the Endolite-foot than with the Sach-foot. Comparing the other prosthetic feet in this way, it shows a lesser evident tendency concerning the maximal kneeflexion.

Muscle activity of the upperleg musculature

To assessment the upperleg musculature, the muscle activity of the m. rectus femoris and the m. semitendinosus has been studied. M. rectus femoris: In normal persons the activity of the m. rectus femoris shows two bursts (Shiavi, 1985; Winter, 1987). The first and biggest burst takes place between ± 90% via 0% up to 20% of the gait cycle. This burst represents the quadriceps activity which is necessary to extend the knee just before heelstrike and to keep a check control on the knee which disposed to bend because of the weight of the upper part of the body. The second burst occurs on the moment of toe-off.

Winters (1987) theory is that this burst represents two functions:

1. starting the hip flexion of the swing leg.
2. preventing the lower leg from staying behind while swinging the upper leg forward.

Maybe there still can be a third explanation, it means the compensation of the bending moment around the knee which arises owing to the push-off just before the toe-off. It is imaginable that the action of the m. triceps surae at the push-off gives cause for the inclination to kneeflexion - also because of the bi-articular action of the both mm. gastrocnemici.

With the users of a legprosthesis we roughly see another pattern in 3 of 5 patients (see fig. 5 as an example). The first burst of muscle activity extends up to 40% of the gait cycle and the second burst is missing. The influence of the type of prosthetic-foot on this pattern is not measurable.

However the influence of the type of shoes on the activation pattern appears to be measurable (see fig. 5). The intensity of the EMG is systematically higher in using leather shoes than using in sportshoes. This occurs in all prosthetic-feet and in all patients. It points strongly to the trend of a systematically higher load of the m. quadriceps in the early stance-phase using leather shoes. The size of this increase varies from 50 up to 100%.

M. Semitendinosus

The activation patterns of this muscle differ strongly per patient. Also because of the bad reproducibility of the EMG-measurement of the m. semitendinosus, the patterns make comparisons with normal persons impossible.

DISCUSSION

Time, stance and symmetry parameters

The time- and stance parameters give little more information. They indicate what was already known. There were good walkers because they had been selected on this criterion. The fact that one can't distinguish a systematic influence of the prosthetic foot on the parameters measured, doesn't mean that there would be no difference . The parameters are determined over a distance of 10 meters. It may be possible that the influence of the prosthetic foot becomes clear after walking over a longer distance or it would be, that the user of a prosthetic foot notices yet a difference while he is sporting. The conclusion only might be, that the type of prosthetic foot does not have much influence on the walking parameters, walking 10 meters on a roughly smooth surface. If one wants to give a statement about other functional circumstances like sport or go upstairs and downstairs, than you will also have to registrate under that conditions. Neither an influence of the prosthetic feet on symmetry is to prove in walking over 10 meters. So it is dangerous to draw general conclusions.

Ankle/knee movement and muscle activity - an integral analysis

It was already established, that the combination prosthetic foot-leather shoes gives systematically more plantar flexion and less dorsal flexion than the combination prosthetic foot-sportshoes. The chance that the systematics as certained are a coincidence is less than equal to 5%. So it is self- evident to search here for an explanation.

The goniometer, with which the ankle angle is measured, is fixed as well to the prosthetic-tube as to a fixed point on the shoe, marked with a sticker. The axis of the goniometer is situated at the level of the imaginary ankle joint, approximately 3 cm. above the upperedge of the shoe and in the middle of the prosthetic-tube.

In outline the attachment of the goniometer has been presented in fig. 4. A difference in heelheight could declare these systematic differences. For, when the prosthetic-tube should take on invariable spatial position with respect to the ground at the moment of the maximal plantar flexion of the prosthetic-foot with both types of shoes, the prosthetic-foot/shoe with the highest heel will have to show directly after the heelstrike a bigger angle alpha with the prosthetic-foot flat on the ground than in case of the shoe (sport) with a lower heel. Likewise, the dorsal flexion angle of the shoe with highest heel will be smaller just before the toe-off than with the lower heel. The systematic differences in fig. 1 and 2 cohering with the difference in the types of shoes, should not to be ascribed to the difference in stiffness of the shoes. This explanation is affirmed by the photos of the below-knee prosthesis with both types of shoes (see fig. 5a and fig. 5b). Fig. 5a shows the below-knee prosthesis with the sportshoe and fig. 5b that with the leather shoe. One can see that the prosthetic-tube leans approximately 4 degrees backwards with respect to the vertical with the ground, while with leather shoes the tube stands nearly vertical.

Fig. 6 shows that the moment of maximal plantar flexion does not differ much between the various prosthetic-feet and shoes. The moment of the maximal plantar flexion has been plotted along the y-axis measured from the moment of the heelstrike and normalized on the average steptime. Zero and 100% represents heelstrike; toe-off is about 60%. We systematically see here that the maximal plantar flexion with the leather shoe is also reacted a little later than with the sportshoe. This can also be explained from the differe- nce of heelheight. The "ankle" has to turn further before the forefoot is flat on the ground as result of the higher heel. This turning (longer) takes a little time.

The stiffnesses of the various prosthetic-feet are reflected in the goniosignals, registrated of the "ankle" of the prosthetic-feet. Nothing has been registrated about a possible mobility of the forefoot, which also works in the biomechanics of walking, presumable especially in the late stance- phase. The meaning of these registrations is extremely relative. Comparing of these gonio-registrations with those of normal persons on behalf of the interpretation and evaluation of walking with a prosthesis is a procedure, which will not give insight. For insight in the processes of prosthetic walking, it is necessary, that one takes into consideration the cohesion in the movements and forces around the "ankle" and the knee. The explanation of the various measured effects demands an integral biomechanical approach, with which the ankle mechanism, the knee-movement and the muscle activity on various places of the leg have to be describe.

An attempt to such an approach can start with the analysis of the knee-movement. One can see in table 1. that the maximal kneeflexion in patients is systematically smaller during stance-phase than in normal persons. The question obtrudes upon why this happens. An other phenomenon without apparently any connection is the systematic effect of the higher heel of the leather shoe on as well the kneeflexion during stance-phase - this becomes longer - as also on the activity of the m. rectus femoris (representatively for the m. quadriceps) which becomes also higher. This put up to thinking. The kneeflexion and the quadriceps load are strongly coupled together. It could be, that the smaller kneeflexion during the stance-phase in patients compared to normals is a mechanism of compensation of an increase of the quadriceps load as a result of the biomechanics changed at the level of the lower leg.

Now it is clear, that there are significant differences in the biomechanical construction of the lower leg of normal persons and users of a prosthesis. First of all the plantar flexors of the ankle lack totally in patients with below-knee amputation. In addition there is not an ankle joint, so that the (prosthetic) foot can't plantarflex directly after heelstrike, till the sole of the shoe rests on the ground. In normal persons the calfmuscles become already active before the end of the double stance-phase (about 10% of the gait cycle). These activity increases, when stance-phase continues. This means that the foot which is flat on the ground, serves as kind of anchor and that a force around the ankle will operate against the dorsal flexion movement of the ankle as result of that muscle activity. This movement is inhibited a little. At that moment the upper body has a forward speed. One can imagine that the inhibition of the ankle movement by the calfmuscles, in combination with the forward movement of the slow mass of the upper leg, trunk, head, arms and swingleg, results in an extending tendency around the knee. By applying a little counter-pressure of the calfmuscles in normal persons the quadriceps load would be reduced during the early stance-phase.

The mechanism described is missing in users of a prosthesis. The result should be at a kneeflexion pattern; comparing it with that of normal persons, that the load of the knee-extensors is much bigger. A way to reduce the load of the knee-extensors is a reduction of the kneeflexion, it means that then the line of gravity of the body minus the lower leg is less far behind the kneejoint. This leads to lower load of the extensors. At the same time less kneeflexion with the same step-time would mean, that the point of gravity of the upperbody would move a little more on high, during stance-phase. This costs more energy. That can be compensated reducing the step-size. If the stepp-size of the patients in relation to theri bodylength is actually smaller than normal, is not examined. With this all in mind the following question comes up: Does the Endolite foot, which gives less kneeflexion than the Sach-foot, reall result in less quadriceps-load and less activity in the m. rectus femoris? Unfortunately it is impossible to answer this question as result of this experiment. The EMG- patterns registrated are badly to compare because of the protocol used. The non-reproducibility, which is introduced by replacing of the electrodes with an interal of one week, makes it debatable to compare these patterns.

Concerning the missing of the second burst in the EMG of the m. rectus femoris: When the m. quadriceps really has to keep the knee around the moment of the toe-off under control, to prevent the knee from bending as result of the calfmusculature-activity, it is not surprising that the m. quadriceps of the users of a prosthesis is not active around the toe-off moment: it is not necessary to compensate for calfmuscle-action. Remark that this analysis has been limited to the mechanics of the affected side. One must not forget that also the contra-lateral side still offers possibilities of compensation, for example: 80% of the propulsion energy of the body comes from the calfmuscles (Winter, 1988). If these calfmuscles are missing on one side, that propulsion must come from another mechanism.

It is possible to strongly contract the hip-extensors of the affected side in the early stance-phase (Winter, 1988). The result could be that now the m. quadriceps also is loaded more strongly in the early stance-phase, it means, less kneeflexion during the early stance-phase (and so on). It is also possible that the healthy side will supply more capacity.

In the above mentioned it has been suggested that a changing biomechanics of the lower leg of the users of a prosthesis will provoke mechanisms of compensation which manifest in muscle-loads and movement patterns modified with respect to normal persons. The exact relations in this process are not yet clear yet. In this connection one can put the case, that, insight in the relations is in dispensable before one can try to improve the "prosthesis- patient" system. The opinion among clinicions, that the optimal walking pattern of users of a prosthesis is reached when it resembles a normal pattern as much as possible f.i. (symmetry of gait) is not based on any analysis of the mechanics mentioned. The lack of such a rational foundation is a strong impediment in the process of improving functional aids like prostheses. A well defined study of the biomechanics of walking with a prosthesis and of the mechanisms of compensation is a condition necessary to improve prostheses.

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Shiavi R. (1985). Electromyographic patterns in adult locomotion: a comprehensive review. Journal Rehabilitation Research & Development. 22, 85- 98.

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Winter D.A. (1988), Biomechanics of below knee amputee gait. Journal - Biomechanics. 21, 361-367.

FIG. 1.

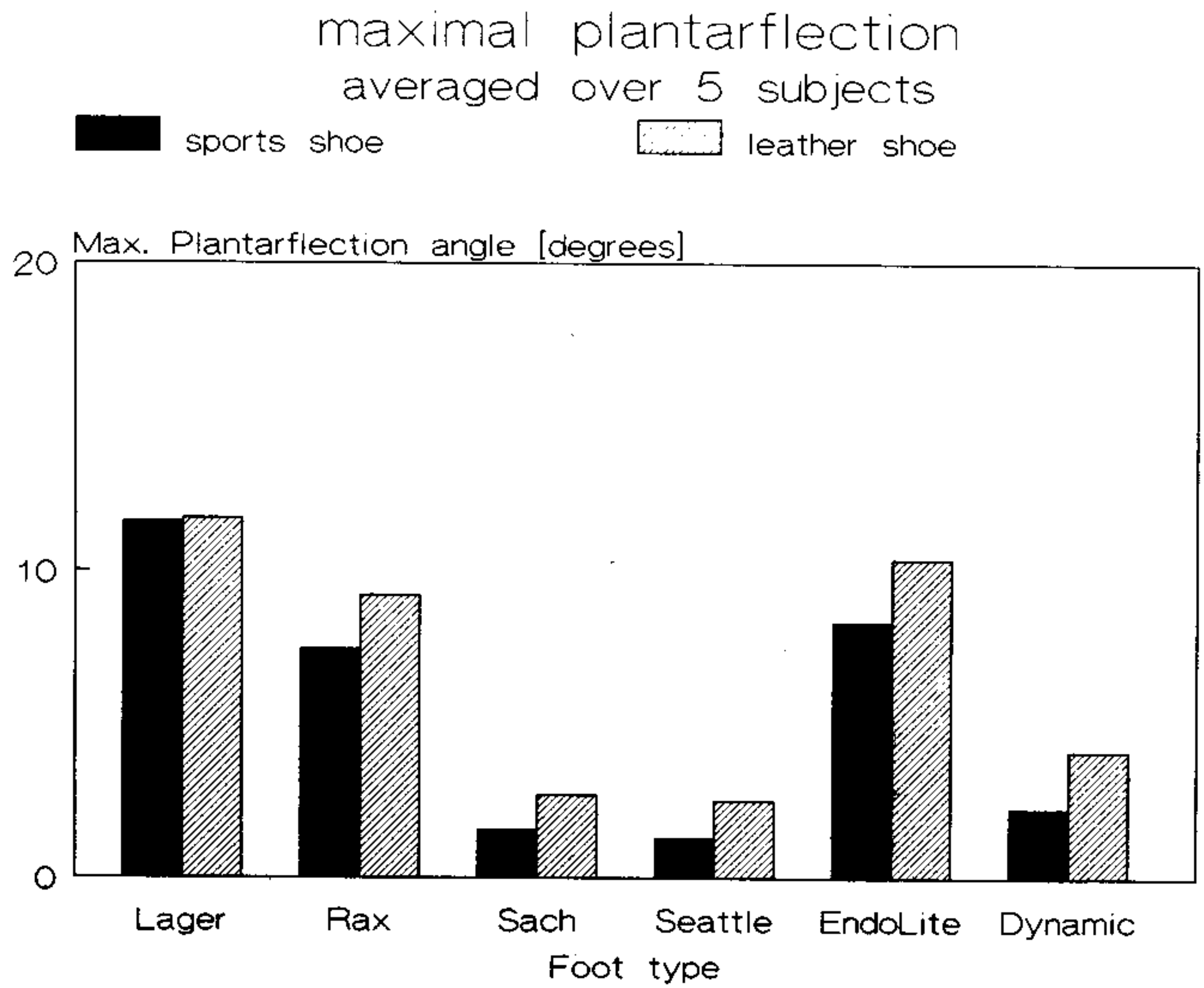

maximal plantarflection
averaged over 5 subjects
sports shoe
leather shoe
Max. Plantarflection angle [degrees]
20
10
0
Lager
Rax
Sach
Seattle
EndoLite
Dynamic
Foot type

FIG. 2.

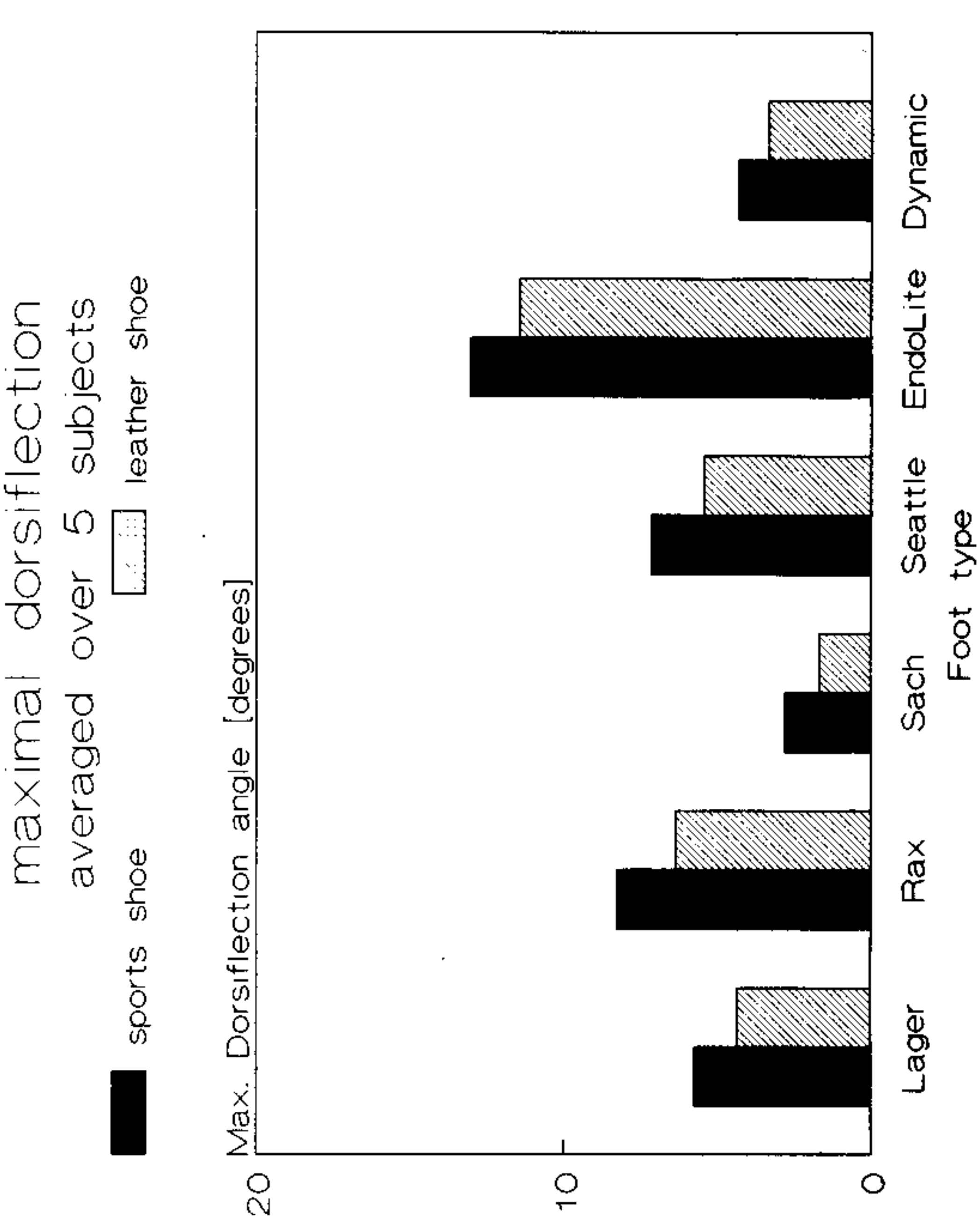

maximal dorsiflection
averaged over 5 subjects
sports shoe
leather shoe
Max. Dorsiflection angle [degrees]
20
10
0
Lager
Rax
Sach
Seattle
EndoLite
Dynamic
Foot type

FIG. 3.

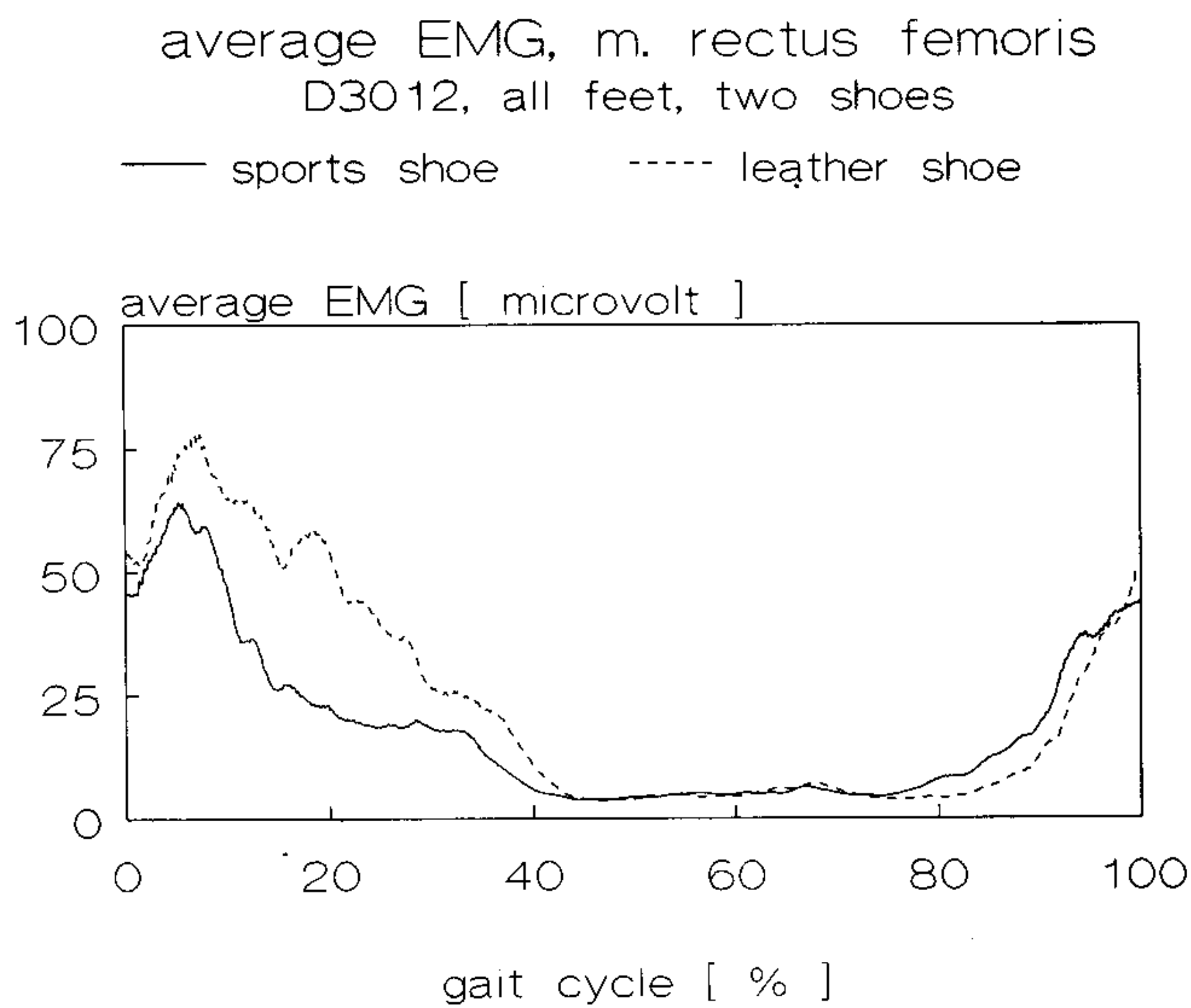
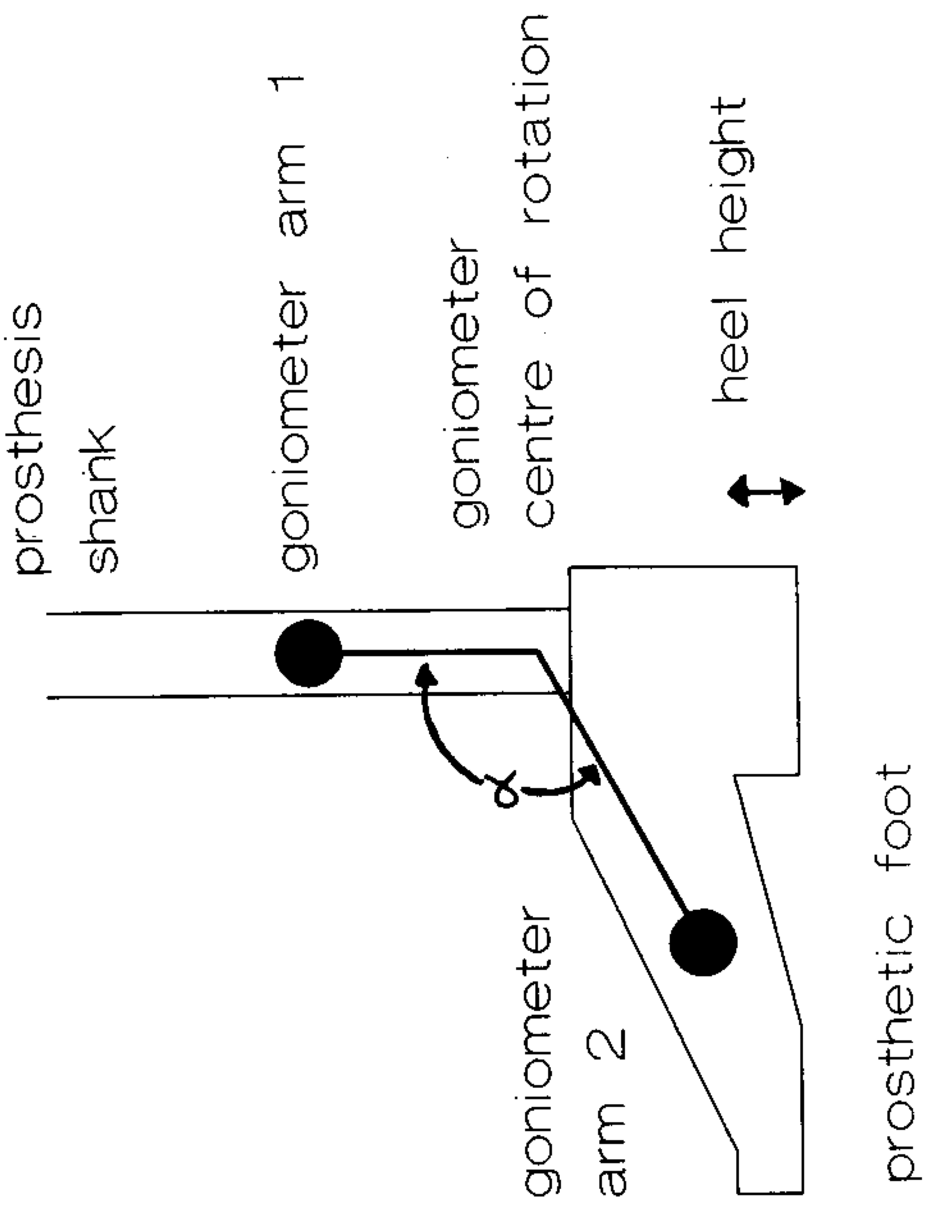

average EMG, m. rectus femoris
D3012, all feet, two shoes
sports shoe
leather shoe
average EMG [microvolt]
100
75
50
25
0
0
20
40
60
80
100
gait cycle [%]

FIG. 4.


prosthesis shank
goniometer arm 1
goniometer centre of rotation
heel height
goniometer arm 2
prosthetic foot
γ

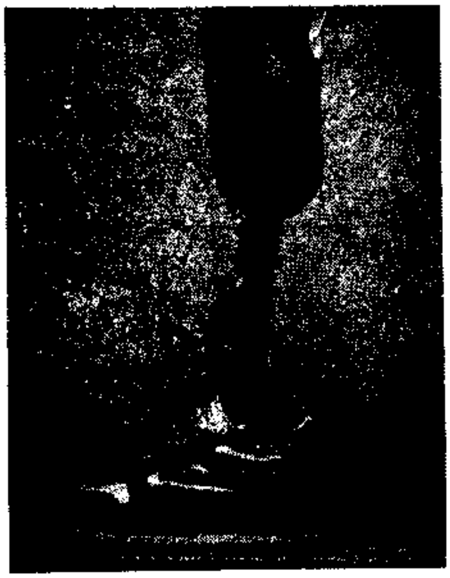


Fig. 5a: below-knee prosthesis with sportshoe.

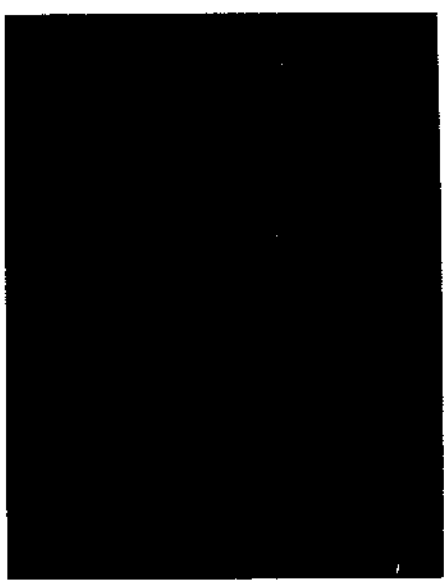


Fig. 5b: below-knee prosthesis with leather shoe.

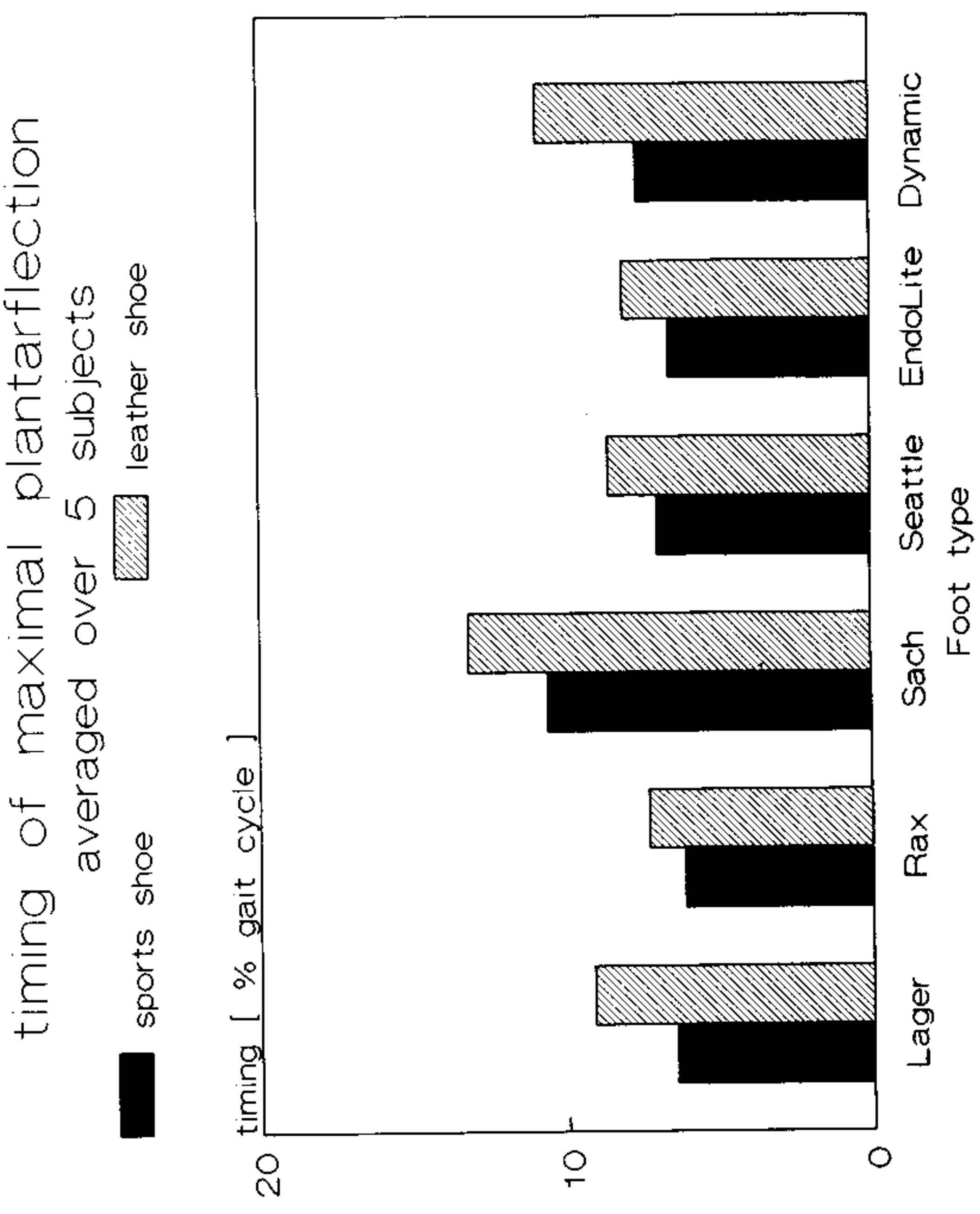

timing of maximal plantarflection
averaged over 5 subjects
sports shoe
leather shoe
timing [% gait cycle]
20
10
0
Lager
Rax
Sach
Seattle
EndoLite
Dynamic
Foot type

FIG. 6.

TABLE 1.

Average maximal knee flexion in stance phase
All feet, leather shoe

subject	average flexion [degrees]	standard deviation [degrees]
D3012	16	3.9
N5704	11.7	1.3
H3704	3.82	0.83
P5912	7.2	1.25
M6604	6.27	1.58

TABLE 2

Maximal knee flexion early stance phase
Maximal flexion with leather shoe
minus maximal flexion with sports shoe, all subjects
[degrees]

Foot	D3012	N5704	H3704	P5912	M6604
Dynamic	3.51	N.A.	-0.26	1.53	-1.75
Lager	3.34	5.99	2.42	6.43	7.38
Sach	6.5	3.5	2.05	3.25	7.351
Rax	-2.06	2.06	2.723	1.82	2.52
Seattle	5.34	3.09	1.11	-0.43	5.89
EndoLite	4.29	1.82	1.63	0.13	4.32

N.A. : Not Available

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