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Effects of sagittal plane prosthetic alignment on standing trans-tibial amputee knee loads

S. BLUMENTRITT*, T. SCHMALZ*, R. JARASCH** and M. SCHNEIDER**

**Department of Research, Otto Bock Orthopaedic Industry, Duderstadt, Germany*
***Orthopaedic workshop, Rehabilitation Centre Göttingen, Germany*

Abstract

The influence of sagittal plane prosthetic alignment changes on loads applied to the ispilateral knee was investigated using 5 trans-tibial amputee subjects. The goal was to determine which prosthetic alignment results in the most energy efficient standing and also minimises stresses on knee structures during standing.

The electromyogram, the external mechanical loading of the prosthetic leg and the amputees' posture were recorded for a wide range of prosthetic alignments. The EMG of the vastus lateralis and biceps femoris muscles was measured bilaterally; the EMG of the gastrocnemius muscle was measured only on the contralateral side. The distance between the anatomical knee centre and each individual's load line, as determined by the Otto Bock "L.A.S.A.R. Posture" alignment system, was used as a measure of the mechanical load applied to the knee joint.

Prosthetic alignment has almost no influence on muscle activity of the contralateral lower limb during static standing. On the other hand, prosthetic alignment has a significant influence on the load applied to the amputee's ipsilateral knee joint. The external knee moments applied to the knee ligaments and knee muscles on the amputated side change systematically in response to different plantar flexion or dorsiflexion angles of the prosthetic ankle-foot. During standing the extensor muscles stabilise the limb by contracting if the load line is located less than 15mm anterior to the anatomical knee centre. The biceps femoris muscle appears to have little or no protective function against hyperextension during standing even if large external knee extension moments are caused by excessive plantar flexion. Such extreme alignments significantly increase the stresses on knee ligaments and the posterior knee capsule.

When prosthetic sagittal plane alignment is altered, the trans-tibial amputee compensates by balancing the upper part of the body over the centre of pressure of the prosthetic foot.

Biomechanically optimal alignment of the trans-tibial prosthesis occurs when the individual load line is approximately 15mm anterior to the anatomical knee centre, permitting a comfortable, energy efficient standing and minimising the mechanical loading on the knee structures.

All correspondence to be addressed to Dr. Siegmar Blumentritt, Otto Bock Orthopaedic Industry Research, Max-Naeder-Strasse 15, 37115 Duderstadt, Germany. Tel: (+49) 5527-8481476 Fax: (+49) 5527-8481484 E-mail: blumentritt@ottobock.de

Introduction

Prosthetic alignment has long been recognised as having a substantial influence on the quality of the trans-tibial (TT) prosthesis. As a prerequisite for daily activities with his prosthesis, the TT amputee must be able to stand comfortably.

Manufacturers' guidelines for static prosthetic alignment are not individualised for each amputee. As a result, prosthetic alignment must be optimised for each individual during iterative dynamic alignment (Pinzur *et al.*, 1995; West, 1987). In current clinical practice, optimisation of prosthetic alignment is a time-consuming, subjective process requiring many years of experience combined with feedback from the amputee for the best result. It is inevitable that this subjective method results in a wide variation in acceptable definitive alignments. Differences in prosthetic alignment have been documented when the alignment procedure is repeated several times, such as when different prosthetists realign the same amputee using identical components (Zahedi *et al.*, 1986; Solomonidis, 1991).

A previous paper (Blumentritt, 1997) reported on the results of posture measurement and prosthetic alignment of 18 experienced TT amputees. The amputee's load line served as an objective, individual reference line. Results of this investigation included:

1. the trans-tibial anatomical knee centre was located between 8 to 40mm posterior to the load line, with a mean value of 18mm posterior to load line;
2. the distance between knee centre and load line was independent of the type of prosthetic foot;
3. definitive alignment of a trans-tibial prosthesis cannot be finalised during one fitting session using the current subjective method.

Breakey (1998) has reported his clinical experience using the Otto Bock "L.A.S.A.R. Posture" system in the alignment of definitive prostheses and has formulated a "Theory of Integrated Balance" using the load line as reference.

It is well accepted that sagittal plane alignment changes to the TT prosthesis will influence amputee comfort during standing. It may be that the acceptance or rejection of a tested prosthetic alignment by the amputee is affected by the muscular or ligamentous forces acting around the knee joint. The goal of this study is to show the results of different sagittal prosthetic alignment on the activity of knee muscles during standing and to propose guidelines for a biomechanically optimal prosthetic alignment.

Methods

The external load on the knee joint of the amputated limb was determined using the Otto Bock "L.A.S.A.R. Posture" alignment system. The posture of the amputee was also recorded using this device.

The "L.A.S.A.R. Posture" device determines the vertical component of the ground reaction forces acting on its sensing platform. When both feet are on the force platform, the patient's weight and the location of the *weight bearing line* can be measured. If only one side, e.g. the prosthetic limb, is on the platform then the force on that leg only and the resultant *load line* will be measured. The horizontal distance between selected anatomical reference points and the load line can be determined be means of this apparatus (Blumentritt, 1997; Breakey, 1998).

The sagittal position of the anatomical centre of the ipsilateral knee was determined according to Nietert (1997), and the position transferred to the lateral side of the prosthetic socket and marked.

The electromyogram (EMG) of the biceps femoris and vastus lateralis muscles of the ipsilateral leg, and the biceps femoris, vastus lateralis and gastrocnemius muscles of the contralateral side measured by surface electrodes, were recorded using the MYOSYSTEM 2000 (Noraxon-Neurodata / Vienna, Berlin). The electrodes were located as described by Noraxon's guideline. The mean amplitude of EMG defined by the integrated EMG divided by recording time was used as a measure of myoelectric activity.

Prosthetic alignment was altered by changing the plantar flexion angle of the prosthetic foot at the ankle.

After the foot alignment was altered, the amputee stood with the prosthetic limb on the "L.A.S.A.R. Posture" force plate, with the contralateral leg standing on an adjacent compensatory block whereby both ankle joints were in the same coronal plane. The amputee was asked to load the prosthesis by half of body weight. Once this was accomplished, EMG readings were recorded for 5 seconds. Simultaneously, the sagittal distance from the load line to the ankle adapter, knee centre, greater trochanter and shoulder of the ipsilateral side were measured. The prosthetic alignment was then changed and, after a short break, the parameters were re-measured. The plantar flexion angle was varied randomly. Figure 1 illustrates these measurements.

Patients

Five (5) experienced trans-tibial amputees who had worn a prosthesis for many years and could walk a significant distance were recruited for this investigation. Consequently, amputees with circulatory impairment were not included in the study. All amputees reported their prostheses to be comfortable and that they had

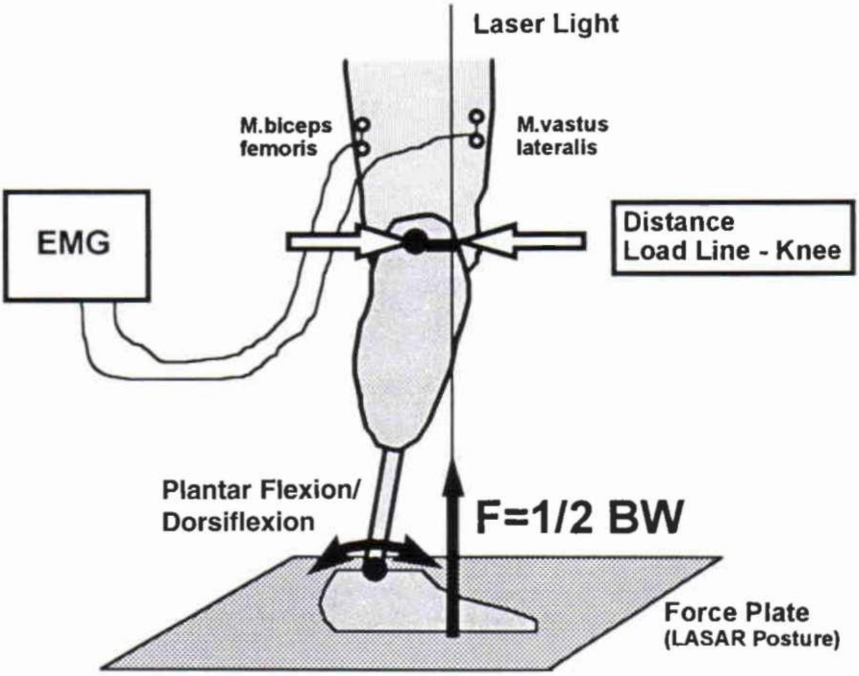


Fig. 1. Measurement principles: The prosthesis is loaded with approximately half of the body weight. By altering the plantar flexion angle of the prosthetic foot, the measured distance between the load line and anatomical knee centre is changed. The EMG of M. vastus lateralis and M. biceps femoris is recorded independently.

no joint pain or range of motion abnormalities.

All subjects wore their customary normal shoes with blocked heels during this study.

Table 1 lists each subject's individual data.

Results

Prosthetic alignment and posture of the amputees

The effect of the flexion angle of the

Table 1. Subject data

Subject	1	2	3	4	5
Age (years)	45	70	42	37	39
Body mass (kg)	83	72	105	61	101
Height (cm)	170	173	183	168	180
Time since amputation (yrs)	26	52	17	34	4
Cause	trauma	trauma	trauma	trauma	disease
Stump length	short	short	medium	long	medium
Side	right	left	right	left	right
Pain in stump	seldom	seldom	no	no	only when stressed
Additional disorders	no	no	no	no	no
Uses cane	no	no	no	no	no
Walking ability (daily)	>5km	>5km	>5km	>5km	300m-1km
Sports activities	no	cycling walking	no	no	no
Prosthesis					
Socket	PTS	PTS	PTS	PTS	PTS
Foot type Otto Bock	1D10	1D25	1D25	1D10	1D10
Size (cm)	26	26	27	24	26

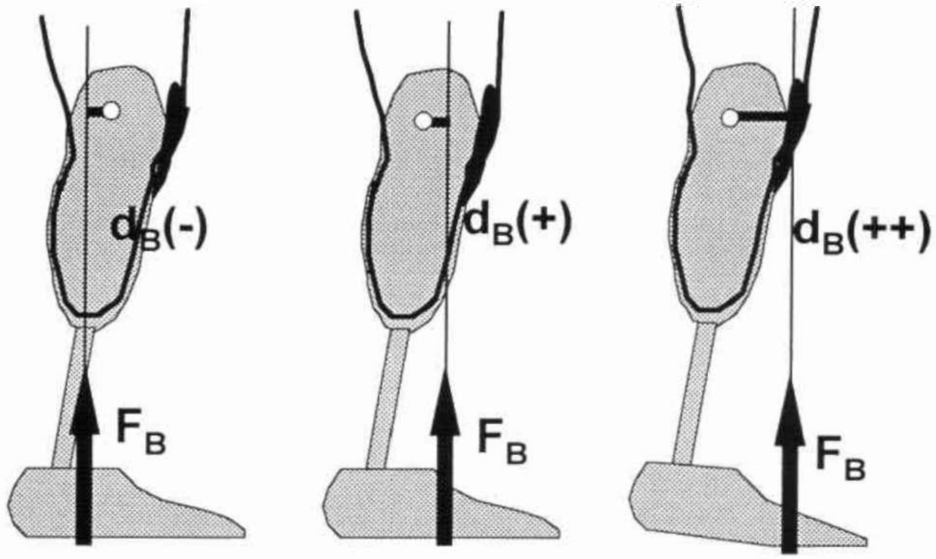
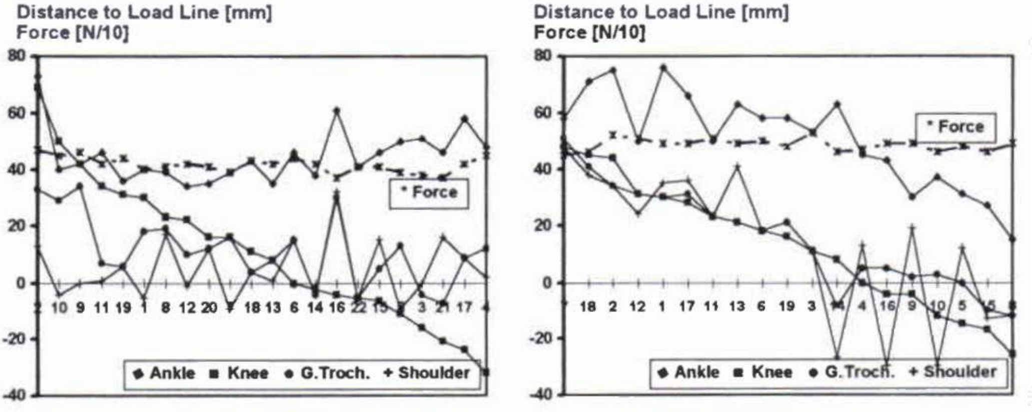


Fig. 2. Effect of plantar flexion of the foot on the mechanical loads applied to the knee joint. The acting lever arm d_B of the ground reaction force F_B changes as the ankle angle is varied (left - dorsiflexed foot, right - plantar flexed foot).

prosthetic foot on the external load on the knee joint was directly visible on all amputees by projecting the laser beam of "L.A.S.A.R. Posture" on the prosthetic side. Increasing the plantar flexion at the ankle tends to move the ground reaction force more anterior to the knee joint. Dorsiflexion of the foot shortens the acting knee lever arm as the ground reaction force falls less anterior to the knee centre. Continued dorsiflexion of the prosthetic foot eventually results in the ground reaction forces acting posterior to the knee centre (Fig. 2). This effect is typical for all amputees tested and was very reproducible.

The amputees compensate for sagittal plane alignment changes by changing their sagittal posture so that the greater trochanter and the shoulder are balanced over the centre of pressure with constant horizontal distances. This supporting point remains almost constant when



a) 1D10 Dynamic Foot

b) 1D25 Dynamic plus Foot

Fig. 3. Load on prosthetic limb (force): distance from ankle adaptor, knee centre, greater trochanter and shoulder to the load line; a) prosthetic foot 1D10 (Pat. 4), b) prosthetic foot 1D25 (Pat. 3). Positive distance means that the selected point is posterior to the load line; negative distance means the anatomical reference is anterior to the load line. The abscissa indicates the test number.

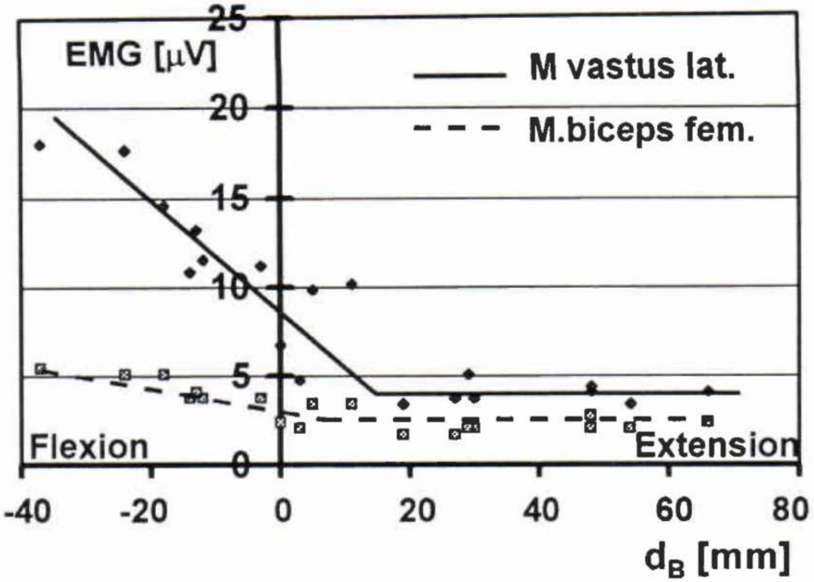


Fig. 4. Mean EMG amplitude of knee muscles on the ipsilateral side (Pat. 5) versus the knee lever arm d_B. The external moment causes knee flexion if the knee lever arm is negative. If the lever arm is positive, the knee joint is subjected to an external extension moment.

flexible keel feet such as the Otto Bock 1D10 dynamic foot are plantarflexed. The centre of pressure varies with the ankle angle for spring keel feet such as the Otto Bock 1D25 dynamic plus design, as shown in Figures 3a and 3b.

Prosthetic alignment and knee muscles' activity

During static standing, the EMG signals of knee muscles and the muscle gastrocnemius of the contralateral leg do not change when the sagittal plane prosthetic alignment is varied. In contrast, the EMG of the muscles of the knee on the amputated side is systematically affected by changes in the ankle angle.

Figure 4 illustrates the typical muscle activity for vastus lateralis and biceps femoris related to the knee lever arm. Once the external knee extension moment is sufficient to fully stabilise the knee passively, the vastus lateralis muscle no longer fires. When the knee extension moment is less, or when the ground reaction force causes a knee flexion moment, the EMG activity of the vastus lateralis increases corresponding to this external moment.

Normally the biceps femoris muscle is inactive during standing regardless of the plantarflexion attitude of the ankle.

Figure 5 shows an overview of the innervation characteristics of the knee muscles of the ipsilateral limb. The regression curves of the

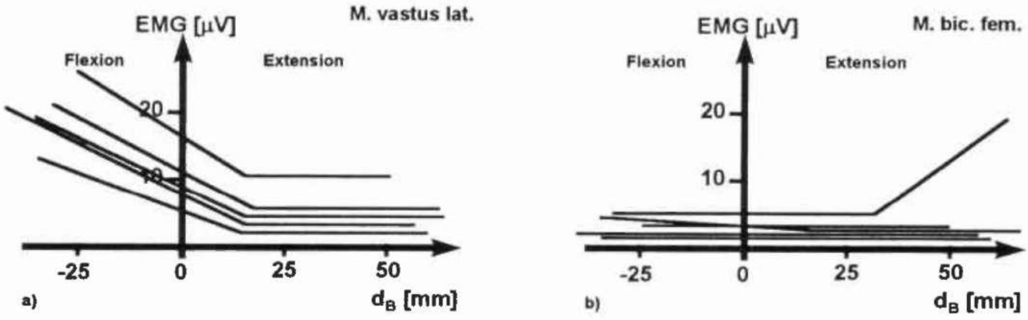


Fig. 5. Linear regression curve of the mean EMG amplitude of knee muscles of the ipsilateral limb versus the knee lever arm (all 5 investigated amputees); a) M. vastus lateralis, b) M. biceps femoris.

mean EMG amplitudes for all 5 amputees verifies that the vastus lateralis muscle is no longer activated whenever the load line is located more than approximately 15mm anterior to the knee centre. If this distance is smaller than 15mm or if the ground reaction force acts posterior to the joint centre, the EMG amplitudes of vastus lateralis increases in direct proportion to the external moment. This correlation is significant and uniform for all amputees.

The biceps femoris muscle was inactive during standing for 4 of the 5 tested amputees. In one case, extreme alignments which created a knee lever arm of more than 30mm triggered the biceps femoris.

Discussion

The importance of prosthetic alignment to the success of prosthetic fitting is well known. Current prosthetic alignment methods require many years of experience from the prosthetist and clear feedback and assessment of the functionality of the prosthesis by the amputee. Thus the procedure of prosthetic alignment is very time-consuming and subjective. Objective, reproducible and faster techniques are desirable.

Visual information about the forces and moments causing the biomechanical function of the prosthesis are one way to make prosthetic alignment more objective. Regularities in prosthetic alignment were seen in prostheses worn successfully over an extended period of time (Blumentritt, 1997). Breakey (1998) has described his extensive experience with 115 trans-tibial amputees and 42 trans-femoral amputees, reporting that when his fittings proved satisfactory over time, the discrepancy between the load line and the body weight line was less than 10mm.

This investigation suggests that the justification for prosthetic alignment can be found by defining a biomechanically sound and energy efficient static posture control.

In this study, the relationship between the electromyographic signal and muscle force produced is directly proportional, because the EMG was measured under isometric conditions. Therefore the EMG value can be replaced by the generated muscle force. The muscle and ligament forces around the knee joint are dependent on the knee lever arm and can be estimated as Figure 6 illustrates.

When the ground reaction force is acting more

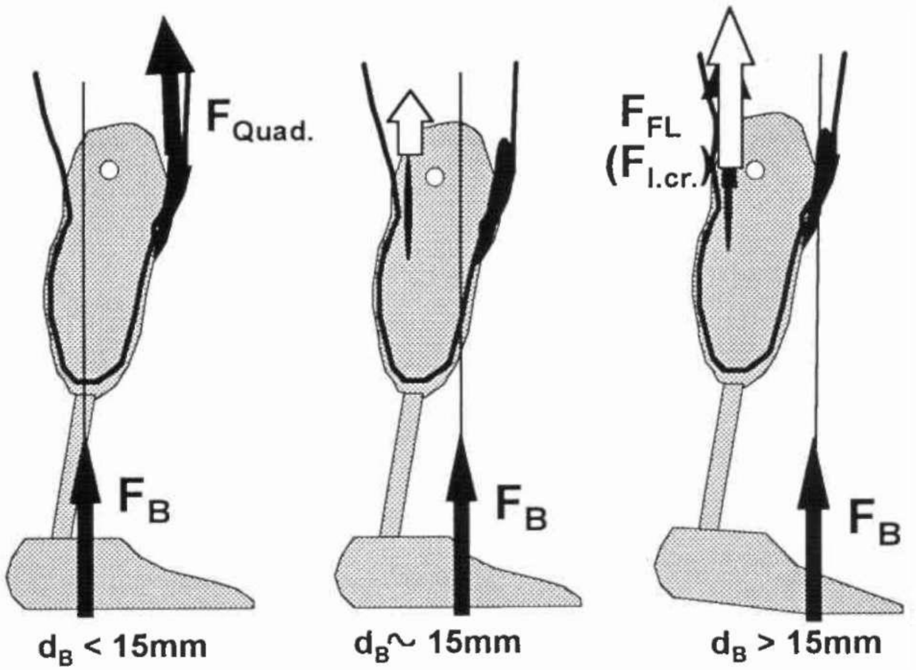


Fig. 6. Diagram illustrating sagittal plane muscle and ligament forces on the knee joint of trans-tibial amputee caused by the external knee load induced by plantar flexion of the foot (left - dorsiflexed foot, right - plantar flexed foot).

than 15mm anterior to the knee centre, knee stabilisation is provided by the ligaments. Prostheses aligned in an equinus position increase the stress on the ligaments and posterior capsule. The knee is hyperextended by such an exceptional external extending moment. This alignment error is presumably avoided in clinical practice by realignment of the prosthesis when the amputee complains. The theoretical muscular protection of the knee joint against high extension load by the muscles which flex the knee seems to be uncommon.

When the ground reaction force acts less than 15mm anterior to the knee centre, the stability of the knee joint will be controlled by using muscular activity of the quadriceps muscle group. The knee joint is extended and the more the load line falls posteriorly to the knee centre the more the knee joint will tend to be flexed. The posterior knee ligament force becomes zero. Such muscular force increases metabolic energy consumption. Stressing the passive structures of the knee joint may result in long term damage, such as that which occurs when the ankle has been fused in equinus to enable the poliomyelitis survivor the chance to walk without bracing on their paralysed leg. Thus, the optimal situation seems to be when the knee centre is located approximately 15mm behind the load line: A minimum of metabolic energy is required while very little extension stress is applied to the ligaments and posterior capsule of the knee.

Individual differences between patients were minimal in the present investigation (Fig. 5). Previously reported individual differences during definitive fittings (Blumentritt, 1997) may have been accidental due to the subjective nature of present alignment techniques. Biomechanically optimal standing alignment can now be defined (Fig. 7).

Use of these guidelines in clinical practice has produced very good fitting results. The amputees clearly describe a noticeable increase in comfort when the prosthesis is aligned as noted above, particularly after using the prosthesis for an extended period of time. This also permits optimal alignment of the prosthesis during only one fitting session.

Of course, the linear position of the foot in the sagittal plane must also be determined. This can be done using conventional visual gait analysis. Sagittal linear alignment is optimal when the amputee slightly flexes the knee on the ipsilateral side during weight acceptance.

It should be noted that these guidelines do not apply to residual limbs with grossly abnormal structures such as occurs when genu recurvatum is present. However, the "L.A.S.A.R. Posture" system may also be useful in such difficult cases by allowing the clinician clearly to visualise the

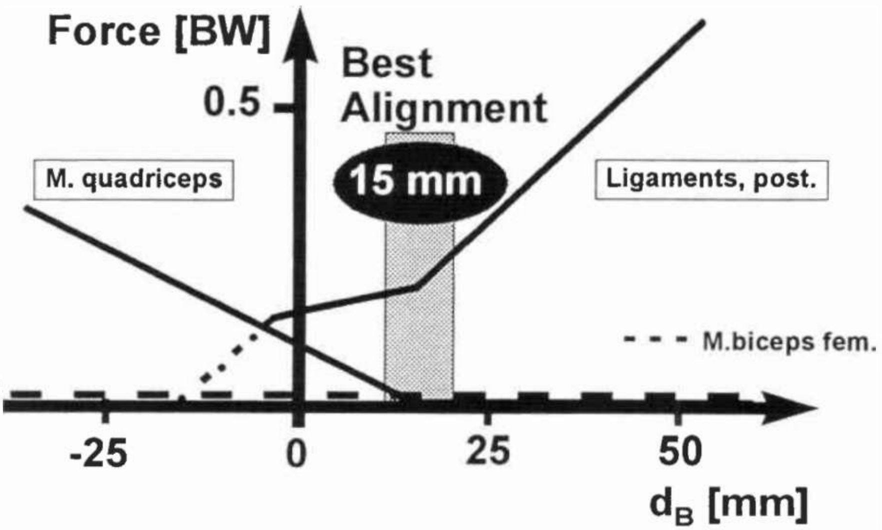


Fig. 7. Force of muscles and ligaments depending on knee lever arm (d_B) of ground reaction force: The external moment causes flexion if the knee lever arm is negative. If the lever arm is positive, the knee joint is loaded with an external extension moment.

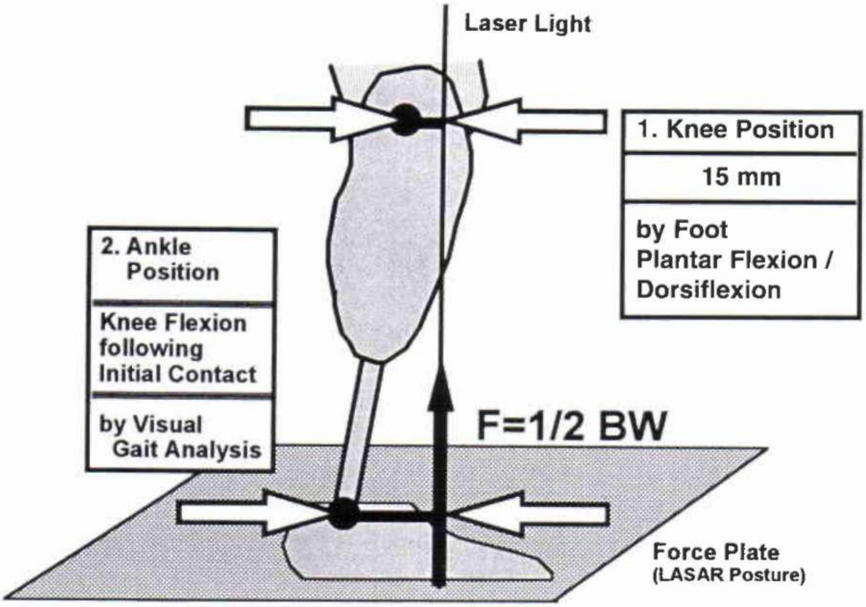


Fig. 8. Static alignment recommendations for trans-tibial prostheses.

vertical ground reaction force during standing.

Conclusions

The present investigation supports the concept that a biomechanically optimal static prosthetic alignment for trans-tibial amputees can be defined. The following principles are proposed (Fig. 8):

1. standing alignment for the trans-tibial amputee is optimal when the ipsilateral anatomical knee centre is 15mm posterior to the individual load line;
2. displacement of individual load line far anterior to the knee centre creates significant mechanical stress on the ligaments and posterior capsule of the knee;
3. when the extension moment on the anatomical knee is insufficient, or when a flexion moment is created, the quadriceps muscles fire to stabilise the joint. This consumes metabolic energy and is undesirable;
4. the static standing posture of the amputee and the activity of the contralateral leg muscles are minimally influenced by variations in the flexion angle of the prosthetic ankle.

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