Prosthetics and Orthotics International, 1983, 7, 41–47

Effects of alignment variables on thigh axial torque during swing phase in AK amputee gait

G. ISHAI, A. BAR and Z. SUSAK*

Julius Silver Institute for Biomedical Engineering, Technion, Haifa, Israel
**Loewenstein Rehabilitation Hospital, Ra'anana, Israel*

Abstract
It is suggested that a major source of discomfort for above-knee amputees during the swing phase of walking, is the thigh axial torque (TAT) transferred at the stump–socket interface. The relation between TAT and variations in its six relevant alignment adjustments, has been investigated. A computerized routine has been established which indicates optimum choice of alignment setting, based on minimal TAT peaks. Feasibility for attenuating swing phase TAT has been demonstrated in three simulated patterns of amputee gait. As a conclusion, it is suggested that a useful clinical tool could be based on the presented alignment optimization procedure and may be expanded to include other factors associated with swing and stance phase comfort and performance.

Introduction

Dynamic effects which occur during the swing phase of walking, of above-knee (AK) amputees frequently cause a sensation of discomfort which is closely related to the thigh axial torque (TAT) transferred at the stump–socket interface, and to the associated shear stresses and angular displacement between the socket and the femur. Time characteristics of the TAT is a function of leg kinematics, mass properties of the prosthetic shank, and of knee-bolt and shank alignment with respect to the socket. The susceptibility of the amputee to TAT during swing phase is higher than during the stance phase due to inertia and gravity loosening the stump socket attachment while the prosthetic leg is swinging forward.

All correspondence to be addressed to Dr. G. Ishai, Julius Silver Institute for Biomedical Engineering, Technion-Israel Institute of Technology, Haifa 32000, Israel.

The kinematics of the prosthetic leg during swing phase is governed by the amputee so as to fulfil functional requirements, such as cadence and ground clearance control. Because his ability to compensate for discomfort associated with TAT is limited, due to loss of voluntary knee control, it is the responsibility of the prosthetist to align the shank and knee-bolt in an optimum manner satisfying both comfort and functional considerations.

New York University (1979) and Radcliffe (1955, 1968), relate to the kinematic phenomenon associated with the TAT, namely, the internal or lateral "whips". Radcliffe states that an artificial leg must usually have the axis of its knee bolt rotated externally about the thigh axis by as much as 5 degrees to compensate for the tendency of the stump to rotate internally during the swing phase and to minimize whip. Two other adjustments which are referred to by Radcliffe (1968) as critical for proper alignment are the amount of abduction of the shank with respect to the socket, affecting both stance phase medio-lateral stability and swing phase TAT, and the inset and outset of the foot which is mainly concerned with whip at the beginning and end of the stance phase. Clinical experience indicates that these guidelines are inapplicable in some complicated cases where final settings are arrived at after many trials. These tire the patient, and moreover, the end result may not fully satisfy the patient and the prosthetist. The source of the problem is mainly due to variability in amputees' gait pattern and in stump musculature as well as the intricate spatial motion of the prosthetic leg, which make achievement of an alignment optimum too complicated without the assistance of quantitative measurements and assessments. Objective measurements via appropriate tools or models may be useful in the clinic only if they determine the magnitude of kinetic variables most closely related to discomfort level and performance in walking, as well as indicate the desired changes in the alignment adjustments.

A previous study (Ishai and Bar, 1981) established a mathematical model, representing the relation between alignment setting and TAT, known to be one of the major causes of discomfort during swing phase. In the present work, TAT characteristics were investigated as a function of gait pattern, and an optimization routine was developed to indicate an alignment setting.

Determination of TAT

Mathematical model

A mathematical model of the swinging prosthetic leg (Ishai and Bar, 1981) determined TAT-time variations. The principle of operation of the model is shown in Figure 1. The values of the alignment variables together with the instantaneous knee flexion angle, determine shank–thigh position at each instant of time. These relative positions together with the measured absolute kinematics of the thigh, are used to determine shank spatial kinematics. The force and moment exerted on the shank by the thigh, at the uppermost point of the shank axis, is calculated using the shank equations of motion. The resultant solution is the torque transferred from the shank to the thigh about the thigh axis (−TAT).

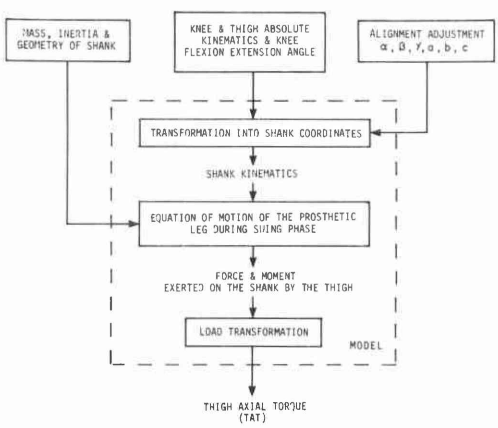

Fig. 1. The mathematical model used to determine thigh axial torque (TAT) characteristics during swing phase.

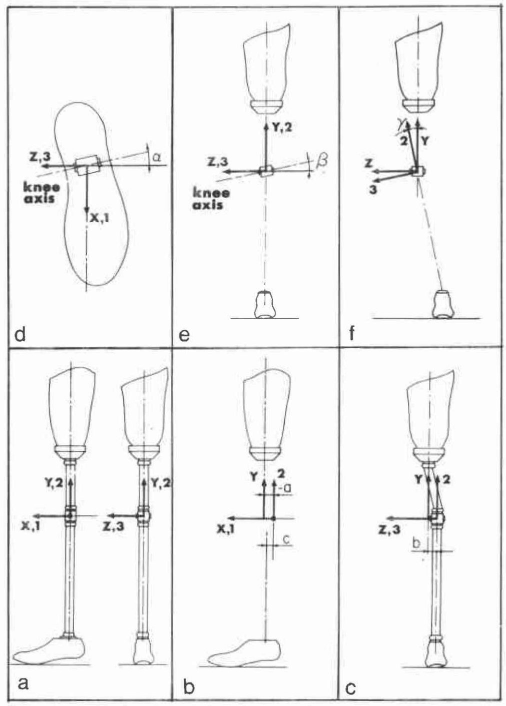

Fig. 2. Definition of alignment variables which affect the swing phase thigh axial torque (TAT).

Alignment variables

A set of six alignment variables has been defined, representing the relevant adjustments which affect the swing-phase TAT (Fig. 2). An $\{X, Y, Z\}$ system of coordinates is attached to the socket of the prosthesis. The Y-axis is the socket's central axis and the Z-axis is parallel to the instantaneous flexion-extension axis of the thigh, referenced to the pelvis. A $\{1, 2, 3\}$ system is attached to the prosthetic shank with the 2-axis coinciding with the shank axis. In the "zero-alignment" state (Fig. 2a), where all alignment variables are set to zero, the two systems of coordinates are coincident.

The alignment variables are:

a and *b* which define knee centre location in the X–Z plane: *a* represents the location along the X and *b*, along the −Z direction (Figs. 2b and 2c)

α and β which define the orientation of the knee bolt relative to the thigh system (Figs. 2d and 2e): α represents the external rotation angle of the knee-bolt about the Y-axis, and β the angular inclination of the knee-bolt with relation to the X–Z plane.

c and γ define the position of the shank with respect to the thigh system (Figs. 2b and 2f): the c adjustment is achieved by shifting the shank forward with respect to the knee bolt. The angle γ is formed by the shank axis relative to the X–Y plane while standing ($\theta_K = 0$).

It should be noted that some alignment systems do not allow the adjustment of all the 6 parameters mentioned, and that each alignment parameter does not necessarily correspond to a single degree of freedom in the alignment device.

Leg kinematics and mass properties of the prosthesis

Due to the unavailability of spatial kinematic data of AK amputees' gait, the kinematic inputs introduced to the model were based on normal gait data published by Lamoreux (1971). In cases where amputees' typical gait patterns were investigated, the normal data were modified: to simulate a case where the amputee has an abducted stump, a constant abduction was added to the normal abduction–adduction angle between thigh and pelvis, at each instant of time. The simulation of circumducted gait was carried out by multiplying the normal thigh abduction–adduction data by a constant factor.

Geometrical dimensions and mass properties of the prosthetic shank are similar to a USMC AK prosthesis type U.S. Universal Multiplex Mark V.

TAT optimization

The TAT simulation based on the mathematical model in Figure 1, may be used in the clinic in the search for an optimum alignment setting which causes minimal peak values in TAT characteristic during swing phase. Using measured leg kinematics and mass properties of the prosthetic shank, the mathematical model may be used to estimate the time change of the TAT for all possible combinations of alignment setting. The setting which results in minimal peak values of TAT is chosen as optimum. The foregoing procedure was found to be time consuming and impractical for clinical use, because it demanded excessive computer time to determine TAT at each alignment setting. In the search for a shorter optimization procedure the relation between TAT and the alignment variables was investigated.

The TAT at zero alignment in the normal, abducted and circumducted gait patterns, are shown in Figure 3. TAT sensitivity to unit variations in the alignment variables about the zero alignment state (1 degree in α, β, γ and 1 cm in a, b, c), is depicted in Figure 4, for each of the walking patterns. There is a fundamental difference between the effects on the swing phase TAT of changes in the γ and b adjustments and of changes in α and β. γ and b cause the shank axis to lie outside the X–Y plane during walking. Both the orientation of the shank axis and the distance of the shank centre of mass from this plane remain constant during the whole walking cycle. On the other hand, α and β cause the orientation of the shank axis with relation to the X–Z plane to change as a function of the magnitude of knee flexion. The greater the knee flexion, the larger the angular deviation of the shank axis relative to the X–Z plane. This difference is expressed in TAT characteristics at the beginning and end of swing phase where TAT values are noticeably higher for a change in γ and b as compared to those obtained by changes in α and β. It should be noted that a numerical filter with a 7 Hz cutoff frequency was used to smooth the kinematic information during data processing and may have filtered out the expected high frequency contents at knee full extension impact near the end of the swing phase. It is expected that contributions to TAT of the γ and b deviations from zero adjustment, would be even larger than those obtained by the simulation.

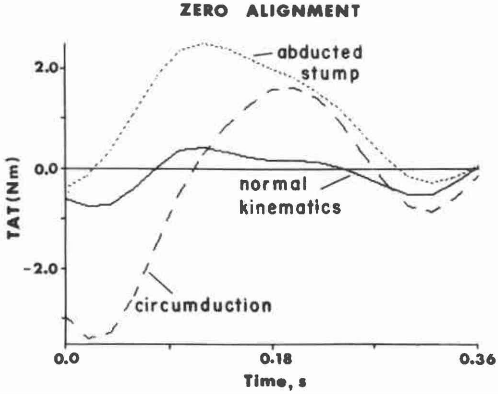

Fig. 3. Thigh axial torque (TAT) during the swing phase in normal gait and in simulated abducted and circumducted amputee gait, at zero-alignment setting.

It is evident that while the sensitivities to α, β, γ and b variations are practically invariant with the investigated walking patterns, the sensitivity of TAT to variations in the a and c adjustments is largely influenced by the investigated gait patterns. This is explained by the a and c adjustments affecting the antero-posterior (X-direction) placement of mass elements of the shank relative to the thigh, thus influencing the magnitude of the moment-arm about the thigh axis, and of medio-lateral (Z-direction) "inertia-forces" acting on the shank. Since the walking patterns mentioned above differ in medio-lateral character of motion, and subsequently in the characteristics of medio-lateral inertia force acting on the shank, the TAT sensitivity to variations in the a and c adjustments is largely affected by the walking patterns. Conversely, since α, β, γ and b adjustments influence the positioning of mass elements of the shank in the Z-direction, the TAT sensitivity to these alignment variables is practically unaffected by the changes in the walking patterns.

TAT sensitivity to alignment variations was investigated in a wide range of simulated alignment adjustments: α, β and γ, -8 to $+8$ degrees,

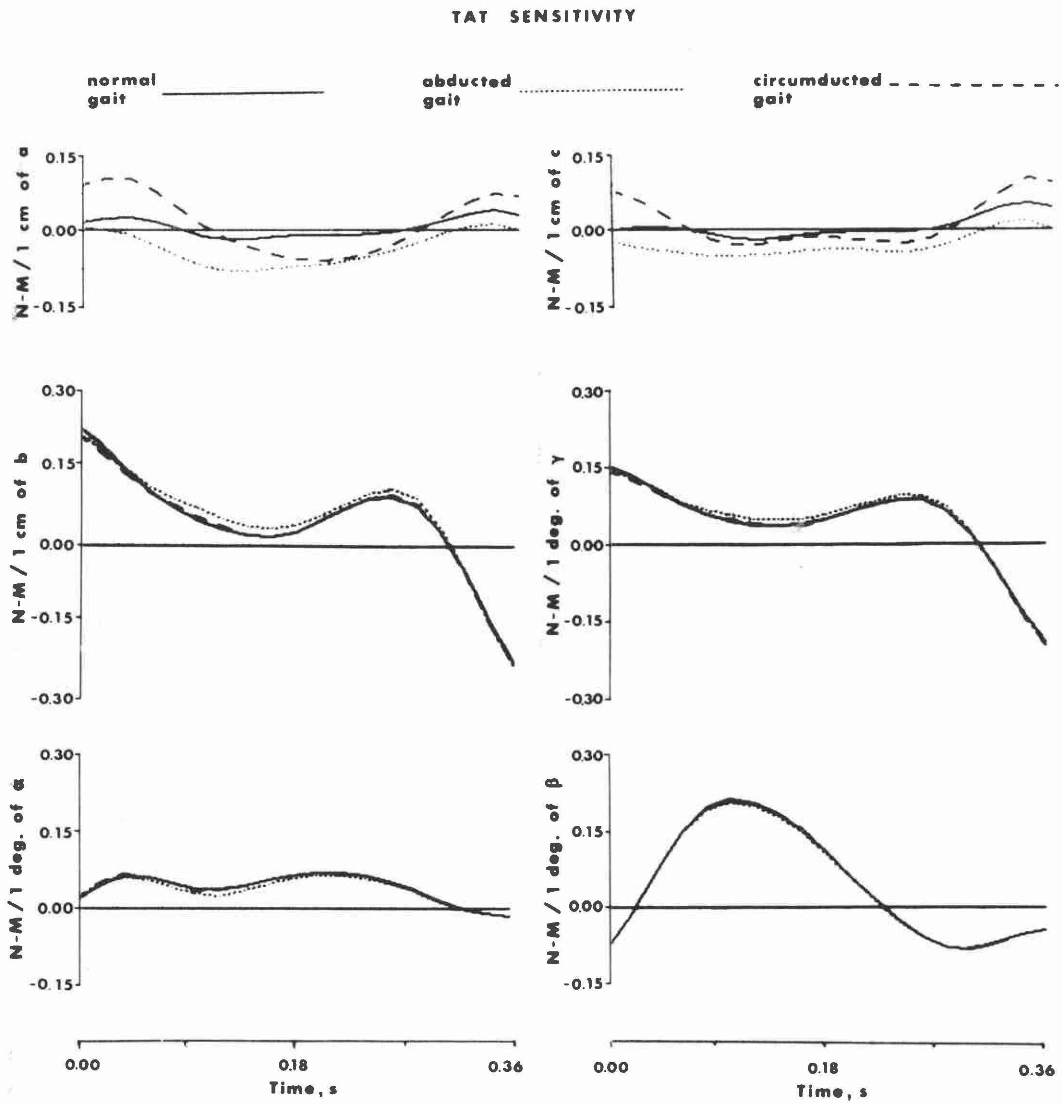

Fig. 4. Thigh axial torque (TAT) sensitivity to unit variations in the alignment variables (1 degree in α, β, γ and 1 cm in a, b, c) about the zero-alignment state.

a and c -5 to $+5$ cm, and b from -8 to $+8$ cm. By introducing unit variations to the alignment variables within these ranges, it was found that deviations of the α, β, γ and b sensitivities from the respective zero-alignment sensitivities, were less than 5%. Conversely, the sensitivity to unit variations in a and c was found to vary significantly within the alignment range. Since the values of the a and c adjustments are usually predetermined from stance phase considerations (as they play an important role in achieving knee stability), their values may not be changed in the optimization of swing-phase TAT.

TAT (denoted by T) is a function of time t and of the six alignment variables:

$$T = T(\alpha, \beta, \gamma, a, b, c, t) \qquad (1)$$

The total differential of T for constant t is given by:

$$dT = \frac{\partial T}{\partial \alpha} d\alpha + \frac{\partial T}{\partial \beta} d\beta + \frac{\partial T}{\partial \gamma} d\gamma + \frac{\partial T}{\partial a} da + \frac{\partial T}{\partial b} db + \frac{\partial T}{\partial c} dc \qquad (2)$$

Since the sensitivity of TAT to variations in α, β, γ and b is practically constant throughout the alignment range and as a and c are constrained by stance–phase considerations and are kept constant in the optimization process:

$$dT(t) = S_\alpha(t)\,d\alpha + S_\beta(t)\,d\beta + S_\gamma(t)\,d\gamma + S_b(t)\,db \qquad (3)$$

where $S_\alpha(t)$, etc., denote the sensitivity characteristics of TAT to unit variations in α, β, γ and b, which are independent of the values of the alignment variables. The TAT, therefore, may be determined by:

$$T(\alpha', \beta', \gamma', a, b', c, t) = T(\alpha, \beta, \gamma, a, b, c, t) + S_\alpha(t).\Delta\alpha + S_\beta(t).\Delta\beta + S_\gamma(t).\Delta\gamma + S_b(t).\Delta b \qquad (4)$$

The meaning of equation 4 is that the TAT for any combination of the alignment variables $(\alpha', \beta', \gamma', a, b', c)$ may be estimated from a given TAT and its sensitivities at another setting $(\alpha, \beta, \gamma, a, b, c)$, providing that a and c are kept the same at the two settings. This equation replaces the mathematical model of Figure 1 in the iterative process of TAT optimization. As a result, the optimization procedure is significantly shortened (only a few minutes), because the lengthy calculations involved in the mathematical model are carried out for just the determination of TAT sensitivities and characteristic at the initial setting.

Results

The TAT-simulation of Figure 1 and the optimization routine of equation 4, were used to test their feasibility in optimizing prosthetic alignment in three cases, each representing a different simulated gait pattern.

Case 1. Normal hip and knee kinematics; 7 cm effective shank valgus

In the conventional alignment procedure, adjustment of γ and b both serve to control the medio-lateral position of ground reaction relative to the hip joint during the stance phase of walking. These control medio-lateral stability and regulate load transfer characteristics to the stump. The result of the b and γ deviations from the zero-adjustment is, therefore, an "effective valgus (or varus)", defined as the resulting medio-lateral shift of the foot from the X–Y plane. An effective valgus may be achieved by an alignment device located immediately above the knee bolt ("lower-thigh device"), or alternatively, by a device located just below the knee ("upper-shank device"). For the former, in addition to shin abduction (γ), the knee bolt is angularly tilted by $\beta = \gamma$; in the latter, the knee bolt remains perpendicular to the thigh axis ($\beta = 0$). For example, an effective valgus of 7 cm may be achieved by a lower-thigh device, having $\beta = \gamma = 7{\cdot}2$ degrees, at a shank length of 55 cm. Stance requirement for knee stability usually dictates a negative setting of a, while the shank axis coincides with the thigh axis at standing, i.e.: $c = -a$. In the present example a and c were $-1{\cdot}5$ and $+1{\cdot}5$ cm, respectively. The resulting TAT, which is depicted in Figure 5, shows peak values of $+2{\cdot}31$ and $-1{\cdot}61$ N-m. The same effective valgus may be obtained by setting $\beta = 0$ and $\gamma = 7{\cdot}2$ degrees, using an upper-shank device. It is clear from Figure 5 that the upper-shank device is preferred, as it gives a markedly lower positive peak value of 0·82 N-m (in the direction of outward rotation about the thigh axis). Using the optimization routine based on equation 4, the setting of $\alpha = -2$ degrees, $\beta = -7$ degrees, $\gamma = 7{\cdot}2$ degrees and $b = 0{\cdot}0$ cm, gave even lower peaks. In the optimization procedure α, β and γ were varied in 1 degree increments over a range of ± 8 degrees. b was determined by the constraint equation:

$$b = EV - l.\,tg\gamma$$

where EV is the effective valgus of 7 cm, and l the

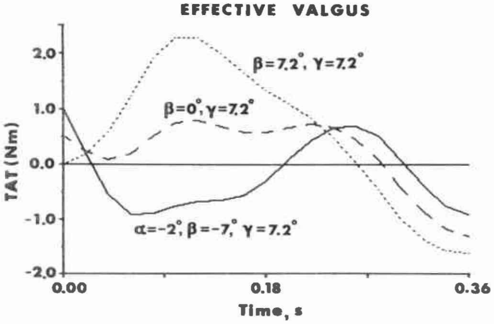

Fig. 5. Thigh axial torque (TAT) during the swing phase, for normal thigh and knee kinematics and with a 7 cm effective valgus of the prosthetic shank. In all settings a and c have the same values as in the reference state, i.e. $a = -1{\cdot}5$ cm and $c = +1{\cdot}5$ cm.

shank length of 55 cm. b was limited to the range of ± 5 cm.

Case 2. Abducted stump

A typical case often met in the clinic requiring special attention, vis-à-vis TAT attenuation, is that of the abducted stump amputee. For example, a 12 degrees of abduction was simulated as previously described where a and c were set to $-1{\cdot}5$ and $+1{\cdot}5$ cm, respectively, to fulfil stance requirements. Figure 6 is the resulting TAT, for the case where all other alignment variables are set to zero. Unfortunately, this setting is usually unacceptable for both stance and swing phase considerations, and it being necessary to medially shift the foot with respect to the thigh. A 7 cm shift results by making $\gamma = -7$ degrees utilizing the upper-shank alignment device. The resulting TAT (Fig. 6), has peak values of $+2{\cdot}2$ and $-1{\cdot}45$ N-m. Using the lower-thigh device, instead of the upper-shank device, to obtain the same medial shift of the foot, i.e., $\beta = \gamma = -7$ degrees, results in a TAT peak reduction to $+1{\cdot}8$ and $-1{\cdot}0$ N-m (18% and 28% attenuation in the positive and negative peaks respectively). Further attenuation of the positive TAT peak to 1·5 N-m, which occurs at the end of the swing phase, is obtained by using optimization routine, as in case 1. Nevertheless, it should be noticed that the further reduction of 14% in TAT peaks (having the $\gamma = -7$ degrees as reference) causes a significant increase of TAT magnitude near mid-swing.

Case 3. Circumducted gait

Another pattern of amputees' gait associated with high TAT magnitudes, is circumducted gait, e.g., the exaggerated medio-lateral movements of the leg in this gait pattern were simulated by multiplying thigh abduction by a factor of 2·0. The resulting TAT (Fig. 7), demonstrates high peak values of $+1{\cdot}8$ and $-3{\cdot}5$ N-m, at the reference setting of $\alpha = \beta = \gamma = b = 0{\cdot}0$, $a = -1{\cdot}5$ cm and $c = +1{\cdot}5$ cm. In this case the high TAT magnitude cannot be significantly reduced by the limited available adjustment range of α, β, γ and b, because of low TAT sensitivity to these parameters. The peak values can be attenuated by shifting the knee centre forward, together with the shank, i.e., by making a more positive and setting c to zero; as depicted in Figure 7, by setting α, β, $\gamma = 0$, $a = 5$ cm, $b = 3$ cm and $c = 0$ cm, a 22% and 34% attenuation is obtained in the positive and negative peaks, respectively. This alignment is unacceptable in commonly used prostheses, because it contradicts the stance phase

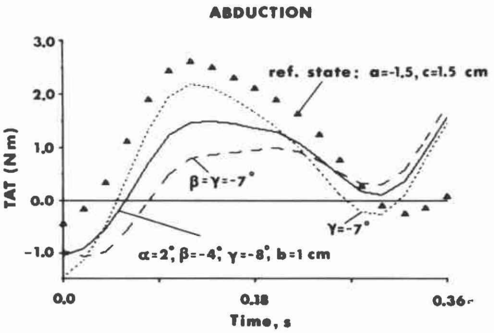

Fig. 6. Thigh axial torque (TAT) during the swing phase, in a simulated gait of an amputee with a 12 degrees abducted stump. In all settings a and c have the same values as in the reference state, i.e. $a = -1{\cdot}5$ cm and $c = +1{\cdot}5$ cm.

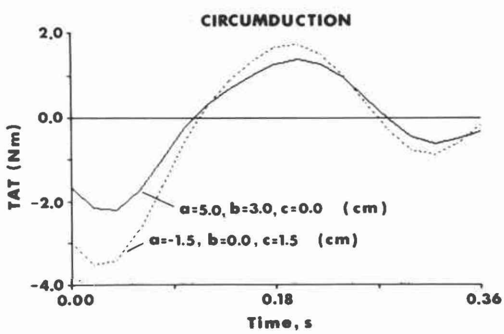

Fig. 7. Thigh axial torque (TAT) during swing phase, in circumducted amputee gait. Circumduction was simulated by multiplying normal thigh abduction angle by a factor of 2. In the two settings $\alpha = \beta = \gamma = 0°$.

stability requirement. However, it may be beneficial for amputees with circumducted gait to have a knee mechanism which allows a positive α-setting. One possibility is a control system which ensures knee locking during the weight-bearing phase, even for a positive α-setting. An alternative is a polycentric knee design having a negative α-setting at full extension which becomes positive with knee flexion.

Discussion and conclusion

1. A mathematical model estimates the thigh axial torque (TAT) during the swing phase of walking, for measured leg kinematics and alignment setting. The model, together with the optimization routine, potentially form a clinically efficient "tool", quickly predicting a final optimum alignment setting.
2. Where determination of TAT may be accomplished by a "pylon transducer" (Berme, 1976) installed in the prosthetic shank and knee goniometer, alignment optimization is made simpler. The reference TAT and the four sensitivity curves (S_α, S_β, S_γ and S_b) are obtained from five successive gait trials. The required changes in alignment are subsequently predicted by the optimization routine, using equation 4. It should be noted that, in this case, the initial alignment setting is not involved in the optimization process and, therefore, need not be measured.
3. The results obtained in cases 1 and 2 indicate that both lower-thigh and upper-shank alignment devices should be used to achieve optimum setting.
4. Because of an absence of rigorous definition of the interaction between TAT and patient comfort, the present paper bases optimum alignment in terms of TAT peak reduction. When defined, an expanded form of optimization criteria, presumably involving parameters such as TAT derivatives and integral, may be incorporated in the optimization procedure.
5. Modified simulated kinematics for normals were introduced to the model of the swinging leg. The optimization procedure will be verified in clinical alignment routines, using measured amputee data in the model. The procedure will be expanded to include other swing and stance phase considerations.

Acknowledgements

The authors wish to thank Mr. E. Lewy from Protesia Lewy Co., Tel-Aviv, for his assistance and advice.

The study was made possible by a grant from the MEP Fund, Women's Division, American Technion Society, N.Y., U.S.A.

REFERENCES

Berme, N., Lawes, P., Solomonidis, S., Paul, J. P. (1976). A shorter pylon transducer for measurement of prosthetic forces and moments during amputee gait. *Eng. Med.*, **4(4)**, 6–8.

Ishai, G., Bar, A. (1981). A computer simulation of leg kinetics for the attenuation of swing-phase thigh-axial-torque in AK prostheses. Internal Report, TSI-06-81, Biomedical Engineering Dept., Technion.

Lamoreux, L. W. (1971). Kinematic measurements in the study of human walking. *Bull. Prosthet. Res.*, **10–15,** 3–84.

New York University Post-Graduate Medical School. Lower-limb prosthetics: [manual]. 1979 revision.

Radcliffe, C. W. (1955). Functional considerations in the fitting of above-knee prostheses. *Artif. Limbs*, **2(1),** 35–60.

Radcliffe, C. W. (1968). Alignment of the above-knee artificial leg. *In* Klopsteg, P. E. and Wilson, P. D. (eds.), Human limbs and their substitutes. Hafner, New York, p. 676–692.