Prosthetics and Orthotics International, 1997, 21, 114–123

The effect of changing the inertia of a trans-tibial dynamic elastic response prosthesis on the kinematics and ground reaction force patterns

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Abstract

The aim of this study was to assess, by means of gait analysis, the effect on the gait of a trans-tibial amputee of altering the mass and the moment of inertia of a dynamic elastic response prosthesis. One male amputee was analysed for four to five walking trials at normal and fast cadences, using the VICON system of motion analysis and an AMTI force plate. The kinematic variables of cadence, swing time, single support time and joint angles for the knee and hip on the affected and intact sides were analysed. The ground reaction force was also analysed. The sample size was limited to one as an example to indicate the changes which are possible through simply changing the inertial characteristics. Descriptive statistics are used to demonstrate these changes. Three mass conditions for the prosthesis were analysed m_1: 1080g; m_2: 1080 + 530g; m_3: 1080 +1460g. The m_1 condition is the mass of the prosthesis with no added weight while m_2 and m_3 were attachments of the same geometrical shape but were made from different materials. It was felt that the large mass range would highlight biomechanical adjustments as a result of its alteration. The effect on selected temporal characteristics were that as the speed increased the cadence changed and the affected side single support times as a percentage of the gait cycle were altered. The effect on the joint angles was also apparent at the hip and knee of both sides. The ground reaction force patterns were similar for all three mass conditions, though the impact peak which was evident in the intact limb was missing, indicating a shock absorbing property in the prosthesis. Clearly, changing the mass and moment of inertia has an effect on the kinematic variables of gait and should be considered when designing a prosthesis.

Introduction

Trans-tibial amputees generally walk more slowly than their normal counterparts and the timing of the gait cycle for both trans-femoral and trans-tibial amputees is not symmetrical (Winter and Sienko, 1988; Colborne *et al.,* 1992, Barr *et al.*, 1992). Recent developments in prosthetic research have resulted in new devices known as Dynamic Elastic Response (DER) prostheses which are claimed by their manufacturers to have superior performance characteristics in gait. However this is a controversial claim with some studies noting limited biomechanical improvements. Macfarlane *et al.* (1991) observed improved kinematic symmetry in the timing of the gait cycle while others comment on the fact that these prostheses are expensive or do not offer any significant biomechanical or energy conservation benefits to the active amputee (Gitter *et al.,* 1991; Colborne *et al.,* 1992; Anzel *et al.,* 1992-1993; Perry and Shanfield, 1993). There are limited studies on the effect of changing the mass and moment of inertia on this

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type of prosthesis, although it is generally understood that the inertial characteristics of a lower limb prosthesis will affect the gait of the amputee. Simulations on trans-femoral amputee models have indicated that altered inertial characteristics of the prosthesis may affect and contribute to the asymmetries involved in amputee gait (Bach *et al.,* 1993), while biomechanical studies on trans-femoral amputees have indicated that altering the inertial characteristics of the prosthesis will have a significant effect on the gait. This leads to the suggestion that these prostheses should be designed to minimise the distal weighting of the prosthesis (Tashman *et al.,* 1985) with the additional note that simply changing the mass of the prosthesis would not be sufficient to alter the swing period of the shank, rather, the centre of mass must also be moved, with a proximal move resulting in a faster swing in the lower limb. Jans and Bach (1995) advise that in the prescription of trans-tibial (TT) prostheses the inertial characteristics should be considered as the amputees analysed subjectively preferred the lighter mass and walked with increased velocity. In normal cadence walking, the swing phase is basically a passive action due to the pendular effect of the swinging limb but when the cadence is changed then the knee flexors are involved in the flexion of the knee in swing (Gage, 1991). In a study assessing the effect ankle weighting would have on normal subjects, Hale (1990) noted that asymmetrical weighting affected walking with a reduction in the single limb support time, increased swing phase and decreased stance phase for the weighted limb. Inman (1967) observed that trans-tibial amputees need a light prosthesis in early swing, so that with the reduced moment of inertia, the prosthesis could be brought to accelerate into swing easily, but that in late swing, the prosthesis should be heavy and thereby with its increased inertia would increase the acceleration and energy supplied to the system. Research on the affect of changing the inertial characteristics for different cadences is lacking in the literature. This raises the question as to the optimal mass and mass distribution of a prosthesis which may be used for different cadences. The purpose of this investigation was to assess the effect of altering the inertial properties of the prosthesis on the temporal, kinematic and ground reaction force variables of the gait cycle.

Methods

One young male amputee (body mass: 79kg, height: 1.87m, age: 24y) was assessed. An experimental DER prosthesis was fitted to a patellar-tendon-bearing (PTB) socket. This prosthesis allowed for the inertial characteristics of the prosthesis to be altered by adding a weight to the distal foot portion. The original mass of the prosthesis was m_1 1080g. Additional masses of m_2 (530g) and m_3 (1460g) were attached to the prosthesis. They were the same geometrical shape but made from different materials, m_2 was aluminium and m_3 was steel. It was felt that a wide range in the added mass would highlight the biomechanical alterations in the gait of the amputee. The addition of the masses had the effect of moving the centre of the mass of the prosthesis distally from 32.1cm, to 33.11cm for m_2 and 39.58cm for m_3. Since the shape of the prosthesis was such that the centre of gravity (CoG) of the prosthesis was located outside of it, the position of the CoG was estimated using the reaction board technique. The moment of inertia of the prosthesis was also altered from 0.156kgm^2 (m_1) to 0.245kgm^2 (m_2) and 0.53 kgm^2 (m_3). The moment of inertia of the prosthesis wascalculated as follows: the period (T) of the prosthesis was estimated by swinging the prosthesis for 10 periods from an almost frictionless pin at the position of the knee joint axis. One period for each of 10 trials was calculated and the average taken. From this the moment of inertia was calculated from:

$$I = (WhT^2)/(4\pi^2)$$

where I is the moment of inertia about the axis of the prosthesis at the knee joint; W is the weight of the prosthesis in air; h is the distance from the centre of mass to the suspension axis and T is the period of 1 oscillation.

The gait analysis on the subject was carried out using a five camera VICON motion analysis system (Oxford Metrics Ltd. Botley, Oxford, UK) and an AMTI force platform linked to an Etherbox data acquisition system and a host computer. The subject walked at each of a self-selected normal and a fast walking cadence along a 12m walkway. The subject was recorded walking through a 0.65m (X:medial/lateral direction) x 2.45m (Y: anterior/posterior direction) x 1.53m (Z: vertical direction) calibrated volume. A single force plate was located in the centre of the calibrated volume. A trial was deemed to be successful if the subject's foot landed on the plate without any alterations to the gait pattern. Five trials for the affected and five trials for the intact sides were recorded for both cadences. During testing with m_3, fatigue was becoming evident after the collection of 4 trials at fast cadence and testing was terminated. Trials were ensemble averaged to allow comparisons across trials. One stride from heelstrike to the following ipsilateral heelstrike was deemed 100% and the averaging was carried out at the 2% level which has been deemed acceptable for walking trials (Winter, 1987). The subject was allowed 15min to become accustomed to each new mass condition before testing was resumed. The subject was required to wear shorts and reflective markers were placed on the subject's skin on the sacrum, the left and right anterior superior iliac spines, the lateral femoral condyle of the right knee, the right lateral malleolus, the head of the right second metatarsal, along the thigh in line with the greater trochanter and the femoral condyle and on the shank in line with the femoral condyle and the malleolus. On the prosthetic (left) side, markers were placed on the prosthesis lateral to the left femoral condyle, at the estimated position of the malleolus on the lateral distal aspect of the prosthesis, the estimated position of the top of the second metatarsal and along the thigh and shank as for the intact limb. Cadence, stance time, swing time, single support time, joint angles for the knee and hip on the affected and intact sides and the three components of the ground reaction force were analysed. The subject wore the same shoes for all conditions. The ankle was not assessed as the addition of the weights did not affect the prosthesis kinematics at the ankle.

Results

Stride characteristics

The means and standard deviations, in brackets, for the walking speed, cadence, stride length single-limb support and swing periods are presented in Table 1 for normal and fast walking trials.

Altering the inertial parameters of the prosthesis showed minimal effect on self-selected normal walking cadence. However, cadence at the fast condition was observed to decrease with each increase of mass.

The velocity of the subject over the plate at natural cadence varied with the alteration of the mass, with the velocity increasing as the mass increased, while at the fast cadence the velocity was consistent, with little change across the different mass conditions.

The stride increased for the faster walking condition and also increased as the mass was increased.

Single limb support decreased as the mass increased for the affected limb (left) for both cadences and concurrently, the single limb support for the intact limb increased with the increasing mass on the affected side. Single

Table 1. Speed, cadence, stride length and single support time for the three mass conditions walking at normal and fast cadences

	m_1		m_2		m_3	
	Normal (σ)	Fast (σ)	Normal (σ)	Fast (σ)	Normal (σ)	Fast (σ)
Speed (m/s)	1.47 (0.03)	1.86 (0.03)	1.57 (0.04)	1.89 (0.06)	1.61 (0.03)	1.86 (0.04)
Cadence (step/s)	86.5 (0.01)	97.1 (0.01)	87.1 (0.01)	96.0 (0.02)	84.4 (0.02)	93.2 (0.01)
Stride length (mm)	1755 (20)	1912 (33)	1811 (31)	1973 (35)	1909 (32)	2001 (35)
Single support left (% cycle)	40.49 (2.2)	39.81 (2.8)	41.58 (3.1)	38.4 (2.3)	39.62 (5.1)	37.84 (3.9)
Single support right (% cycle)	39.87 (2.6)	41.6 (2.6)	41.35 (2.3)	40.83 (3.1)	42.04 (1.5)	41.19 (5.2)

limb support was asymmetrical for both cadences and the extent of the asymmetry was affected by the mass condition. Using the lightest prosthesis, the intact single limb support was closest to that of the prosthetic single support time for fast cadence, however, for the normal cadence m_2 produced the condition where the single support was closest for both sides.

The graphs of the joint angles in Figures 1 and 2 are for one stride period, from initial contact to initial contact of the same limb. The graphs illustrate the means for the trials collected standard deviation bars are reported in the text for the key instances.

The sagittal plane hip joint angles for the intact and affected limbs for the different mass conditions are shown in Figure 1 for the normal and fast cadences. Knee joint angles are shown for the normal and fast cadences in Figure 2.

Hip joint angle

The affected hip flexion/extension pattern for the different masses was similar. There is between 40° and 50° of flexion at initial contact depending on the mass (m_1: 47° (1.2); m_2: 49° (1.4); m_3: 41° (0.7)). Prior to toe off the hip is in full extension for the lesser mass conditions, m_1 and m_2. For the greatest mass (m_3), the hip is hyperextended to a maximum of 11° (0.8). At mid-swing, the hip showed greatest flexion for m_2, and was most extended for m_3.

The mass also has an effect on the hip angle of the limb and the patterns are similar to the

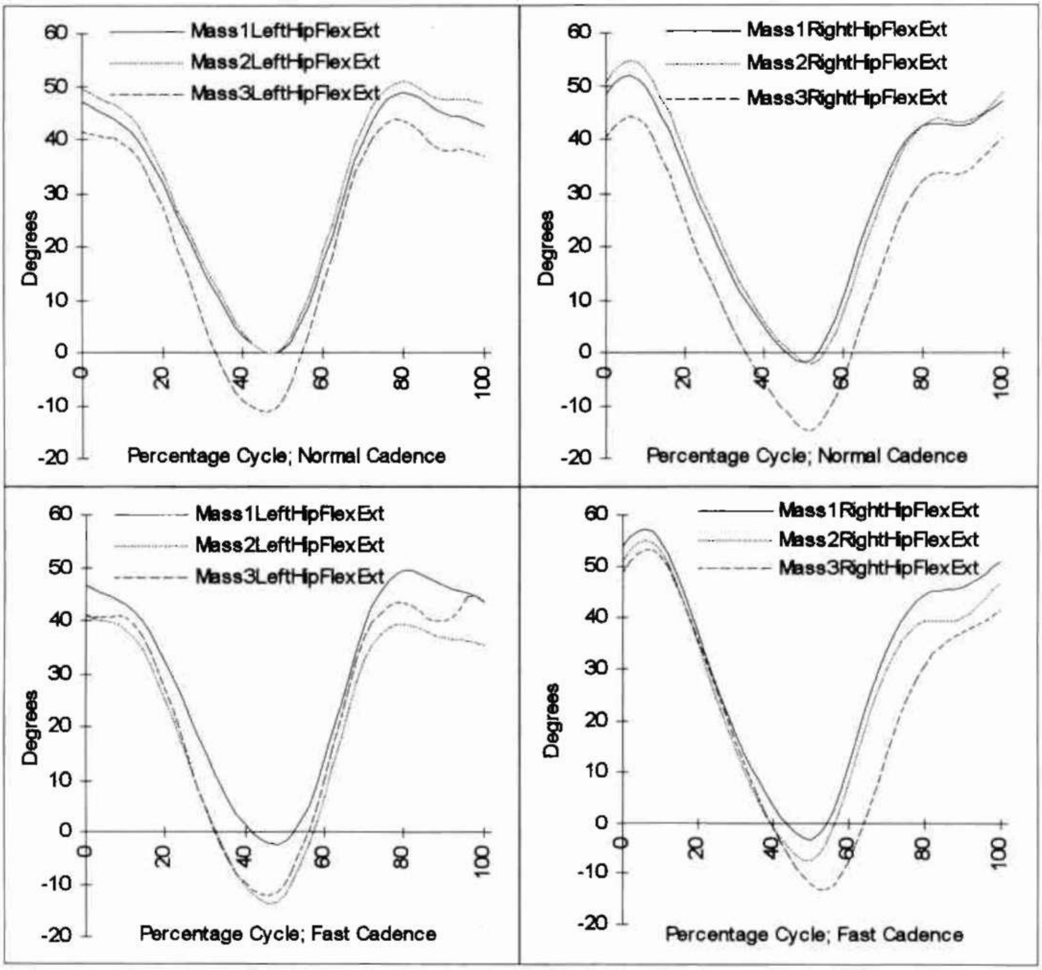

Fig. 1. Affected (left) and intact (right) hip joint angle, normal (top) and fast (bottom) cadences.

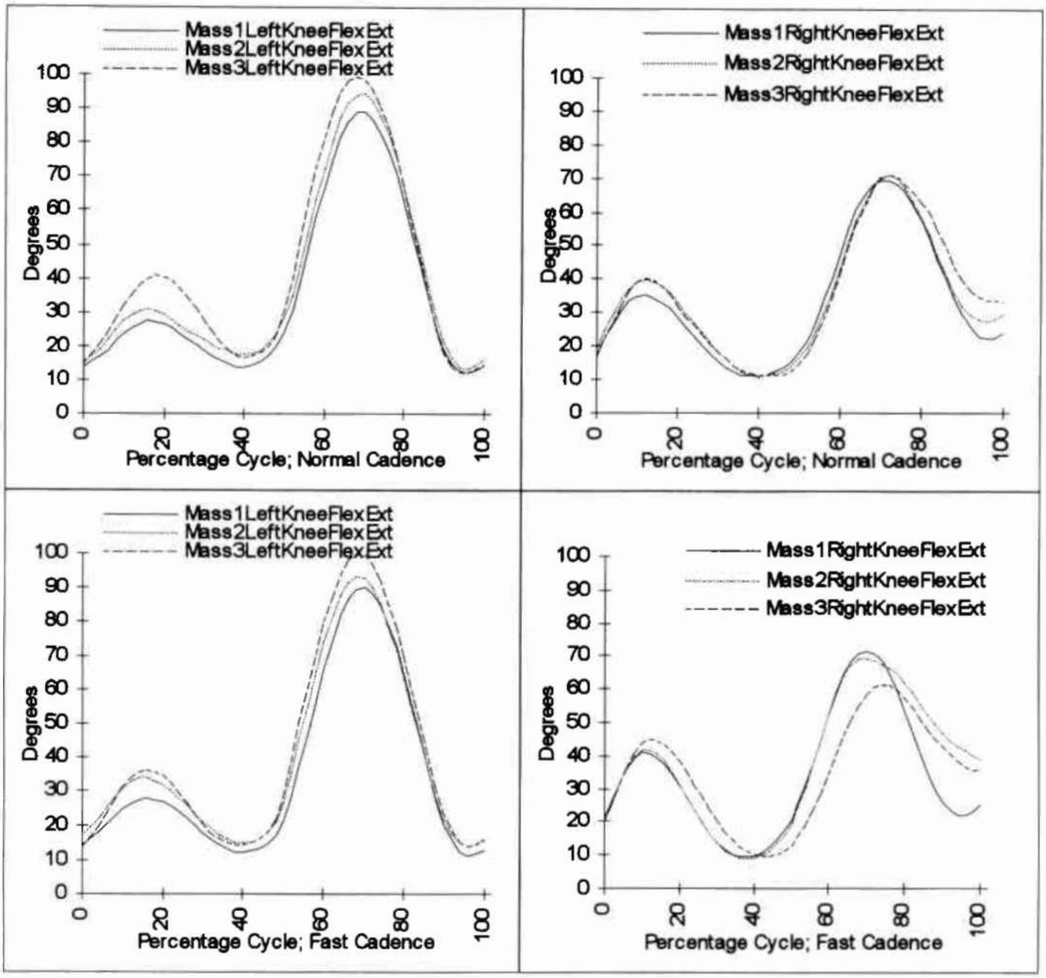

Fig. 2. Affected (left) and intact (right) knee joint angle, normal (top) and fast (bottom) cadeces.

affected limb. At initial contact the hip is flexed to 48° (0.8) for m_1, 50° (0.9) for m_2 and 40° (0.7) for m_3 and unlike the affected limb, the hip continues to flex in early stance before extending at 10% of the cycle. This is probably as a result of the action of lifting, contra-laterally, the prosthesis into swing. Prior to toe off for all three mass conditions the intact hip hyperextends somewhat (2° for m_1 (0.7) and m_2 (1.3)) although it is most extreme in the case of the greatest mass at 15° (1.0). In swing the hip on both the intact and affected sides is more extended for m_3, except for the affected limb at fast cadence where it is more extended for m_2.

Walking at the fast cadence, the affected limb flexion/extension pattern for the hip joint angle is similar to the normal walking results for each of the conditions, though the magnitudes vary. The hip begins to extend immediately after initial contact for m_1, though for m_2. the hip extends more slowly from the initial contact of 40° (1.2) of flexion and begins to extend rapidly at 10% of the cycle. For m_3 the hip remains at 40° (1.4) of flexion until 14% of the cycle. For m_1 the hip continues to 2° (0.8) of hyperextension, and then begins to flex as the body prepares to lift the prosthesis into swing. Maximum hip flexion in mid to late swing is 49° (0.7). For m_2 and m_3 the hip flexion extension pattern is virtually identical until mid to late swing, when the hip becomes more flexed for m_3 than for m_2, except as indicated above for the affected limb at fast cadence.

For the intact limb at fast cadence the pattern is again similar to the pattern at normal cadence, though the mass variation has a different effect. In this instance, the three masses produce similar results in early stance. At initial contact the hip is flexed at 50-51° in all cases. Again, similar to the normal cadence pattern, the hip flexes slightly before extending. Slight discrepancies occur in the timing and degree of maximum extension before the hip begins to flex in preparation for swing. During swing the hip of the intact limb is most flexed for m_1 and least flexed for m_3 when the prosthesis is in stance.

Knee joint angle

The knee joint angle for the prosthetic limb follows a similar pattern to the normal patterns published in the literature. At normal cadence the knee is in 14° (0.7) of flexion for m_1 and 15° (1.0) of flexion for m_2 and m_3. Maximum flexion is early stance for m_1 is 27°, m_2 31° and m_3 40° (1.2). During terminal stance and pre-swing the knee is flexing and this reaches a maximum just after toe off (m_1: 89° (1.4); m_2: 94° (1.8); m_3 100° (0.8). After reaching this maximum, the knee begins to extend in preparation for the next stance period.

At normal cadence the intact knee is not substantially affected by the mass change and follows a normal pattern. The knee is in slight flexion at initial contact (m_1 and m_3: 17° (1.3); m_2: 19° (1.4). The knee flexes through early stance to a maximum of 35° for m_1 and 40° for m_2 and m_3. The knee extends for mid-stance and again flexes in terminal stance and pre-swing in preparation for toe off. After toe off maximum

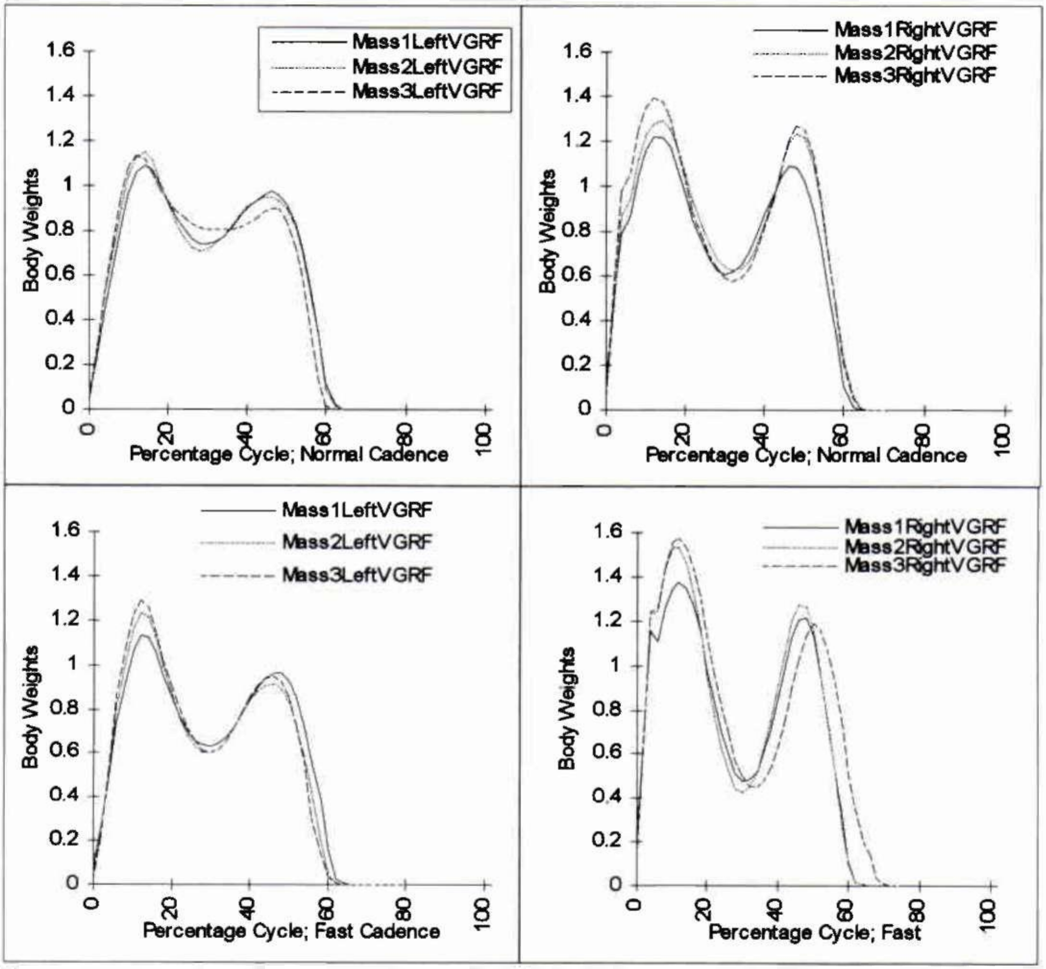

Fig. 3. Affected (left) and intact (right) vertical ground reaction force, normal (top) and fast (bottom) cadences.

flexion is reached (m_1: 69° (0.6); m_2: 71° (0.9); m_3: 71° (0.8)) and again the knee extends to prepare for the oncoming stance.

The fast cadence affects the knee flexion extension pattern for the intact limb. The knee flexes more in early stance when using the greatest mass (45°) and the extension to mid-stance is slower, reaching maximum extension at 48% of the cycle. The flexion is preparation for swing is slower, and the knee does not flex as much for m_3 as for the other two conditions. Throughout the swing period the intact limb is less flexed when the heavier prosthesis is in stance.

Ground reaction force patterns

The vertical ground reaction force (VGRF) and anterior-posterior ground reaction force (A-PGRF) are shown in Figures 3 and 4 for normal and fast cadence. The two vertical peaks typical in walking exist though the second active peak is much lower than normal, but is nonetheless typical for trans-tibial amputees.

The VGRF data is similar for all the mass conditions for the affected limb. The first active peak as the body accepts weight on the supporting affected limb is slightly higher for m_2 at 1.15BW (0.05) than for m_3 at 1.13BW (0.01) and m_1 at 1.09BW (0.01). The trough between the 2 peaks is similar for m_1 and m_2 (0.74BW, 0.71BW) and the curve rises to the second peak values which are below normal (0.97BW, 0.95BW, 0.90BW).

Under these conditions a noticeable impact peak exists for all conditions on the intact side at initial contact. The intact side first active

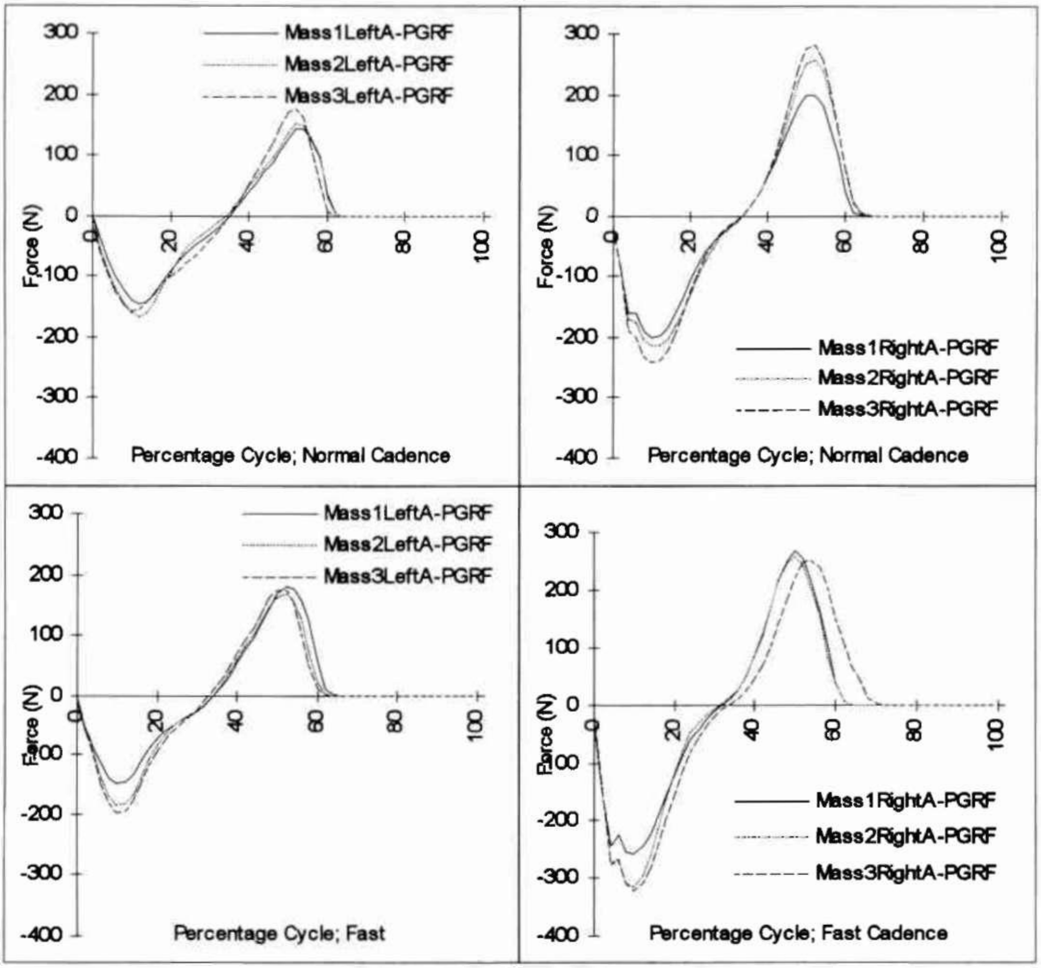

Fig. 4. Affected (left) and intact (right) anterior-posterior ground reaction force, normal (top) and fast (bottom) cadences.

peak is related in magnitude to the mass condition and is highest for m_3 (1.39BW) and lowest for m_1 (1.15BW). The trough is similar for all 3 mass conditions, as is the second peak.

Walking at fast cadence, on the affected limb, there is little alteration to the magnitudes of the first peak compared to the normal cadence, except for m_3 which has increased to 1.29BW. The trough is lower at the faster cadence as the body moves over the limb. The second peak is the same for all three mass conditions and similar to the force at normal cadence.

For the intact limb, the impact force is pronounced and again the first peak is related to cadence, with a larger magnitude than for normal cadence. The trough is lower than for the affected limb and the second peak is similar for all conditions and for the normal cadence (1.21BW, 1.27BW, 1.19BW)

There is little variation in the A-P GRF for the mass conditions ipsilaterally, however compared to the intact limb the magnitudes differ. In all cases, the braking and propulsive forces on the affected side are smaller than for the intact side.

Discussion

Varying the mass resulted in altered temporal and kinematic characteristics and ground reaction force magnitudes in the trans-tibial amputee's walking gait. This is to be expected as the altered inertial characteristics will affect the prosthesis in swing and also the accelerations at the commencement and conclusion of the affected swing phase. Across prosthetic conditions the self selected normal walking cadence was consistent, though m_2 seemed to result in a slightly faster normal cadence. The lightest prosthesis produced the most rapid cadence. The minimal disturbance in the normal cadence when the mass is altered indicates that the amputee is able to maintain his normal walking cadence over a limited distance and number of trials presumably by adapting other aspects. The reduction in the fast cadence with increased mass is similar to the findings of Jans and Bach (1995) and Hale (1990). The velocity over the plate at normal cadence increased as the mass of the prosthesis was increased and this faster velocity was achieved through a longer stride. Slight variations are apparent in the single support time. At normal cadence the single support time is nearly symmetrical for m_1 and m_2. At the fast cadence the observation that the prosthesis is in single support for a shorter time compared to the period for the intact limb corresponds with findings for normal subjects (Hale, 1990). For the greatest mass, the single support asymmetry is obvious for both cadences and this may be as a result of the amputee feeling insecure.

Hip flexion/extension in most cases follows a similar pattern with the different masses. The heavy mass caused the hip to extend more throughout the gait cycle relative to the other mass conditions. Interestingly, the right (intact) limb is also affected by the mass condition, and this is almost definitely a compensation for the action of the limb, although this compensation is not quantifiable without a joint moment and power analysis. The similar patterns for m_1 and m_2 indicate that the mass difference does not affect the kinematics of the hip of the amputee walking at normal cadence for these masses. However, the hip flexion extension pattern for both limbs at the fast cadence for m_2 was similar to that for m_3. At the fast cadence compensatory mechanisms employed by the amputee to successfully walk are evident at the hip in stance and swing. Throughout swing the hip is more flexed than expected, probably to ensure toe clearance as the prosthesis does not allow any dorsiflexion at mid-swing.

The knee patterns are not affected to the same extent by the mass changes as the hip patterns, although again the differences in the mass conditions are most obvious for the faster cadence. The knee is flexed at initial contact and continues to flex through early stance. This flexion is a result of the amputee preparing for foot-flat. During mid-stance to terminal stance and through to pre-swing knee patterns for the three masses were similar. Walking at his normal cadence, the amputee bent his knee more during swing for m_3 on the affected side.

The normal cadence intact knee pattern is not greatly affected by the mass conditions, indicating that the amputee is able to maintain normal kinematics for these conditions.

At fast cadence the intact leg is affected by the mass alterations, most notably in the timing of the knee pattern for the greater mass. The knee extends and flexes more slowly than for the other two masses both while the prosthesis is preparing for swing and while it is in swing. This is as a result of the weight imbalance caused by the large prosthesis mass of the contralateral limb swinging rapidly, and the knee is involved in ensuring stability at the joint. The swing phase kinematics also indicate that for the heavier prostheses, the knee of the intact limb does not flex to the same extent as for the lighter prosthesis.

The vertical ground reaction force patterns vary for the different prosthesis masses also, although the difference between the intact and the affected limbs is more substantial. Anzel *et al.* (1991) suggest that reducing the load on the intact limb could reduce the effect of any bilateral pathology and possibly also delay the onset of the pathology. However in this study, for all masses, the intact foot struck the forceplate with a higher force than the affected side. The m_1 condition produced the most symmetrical forces at normal walking, although at fast walking this was not the case and the greater masses showed a higher loading force which is more symmetrical. The prosthetic limb showed an increase in the first peak from normal to fast cadence which is expected as the foot will strike the plate with a higher acceleration. The trough of mid-stance is higher for the affected side compared to the intact limb and indicates a higher acceleration of the centre of mass over the prosthesis in stance. The second vertical peak does not increase form normal to fast cadence for the prosthesis under any condition. This indicates the inability of the plantarflexors to contract and therefore the inability of the prosthesis to push off into swing, rather it is merely a support on which the body balances. The impact peak which exists for the intact limb at both cadences is missing for the prosthetic side and indicates that there is shock absorption in the mechanical construction of the prosthesis at initial contact.

The similarity of the A-P GRF for all the mass conditions in normal walking indicates that prosthetic mass does not have an effect on the braking and propulsive forces in walking. Walking fast however results in a larger braking force for both m_2 and m_3.

Conclusion

The results indicate that the inertial characteristics of the DER prosthesis affect the temporal and kinematic gait characteristics of trans-tibial amputee gait.

The joint kinematics are affected by the mass condition with the fast cadence demonstrating the most obvious disturbances when using the heaviest prosthesis. It appears that the joint kinematic disturbances can be kept to a minimum for normal cadence walking using the different mass conditions for the limited period of testing, although at the fast cadence the inertial manipulations result in kinematic and GRF adaptations to the increased loading. Most notably, the hip of both the intact and affected limbs is more extended in swing at the normal cadence and the knee seems to flex as a probable consequence, although using the m_3 prosthesis a number of trials had to be discarded as the prosthesis hit the floor in mid-swing. The intact limb is also more extended at the hip and knee in swing when using the m_3 and this may imply a lack of confidence by the amputee when the heavy prosthesis is in stance. The hip flexion in stance is much greater than for the normal population cited in the literature.

The inertial characteristics should therefore be taken into consideration when designing a new prosthesis and especially if the prosthesis is to be used over a number of cadences. The effect on the gait of the amputee is limited due to the restricted laboratory conditions for the normal cadence variables though the faster cadence indicates that a light prosthesis is more suited to the gait of a trans-tibial amputee.

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The Brian Blatchford Prize has been established by the Blatchford family to honour the memory of Brian Blatchford. It is awarded every three years at the World Congress of the International Society for Prosthetics and Orthotics.

The Prize of £2,500 will be awarded to an individual who has an outstanding record of innovative achievement in the field of prosthetics and/or orthotics. The achievement should be related to prosthetic and/or orthotic hardware, or scientifically based new techniques which result in better prostheses or orthoses. The President, in seeking to identify the recipient of the award, will also consider nominations or applications from National Member Societies or individuals. Such nominations or applications should contain a justification together with a curriculum vitae of the candidate and should reach the President of ISPO by 1 January 1998 at the following address:

Seishi Sawamura,
The Hyogo Rehabilitation Centre,
1070, Akebono-cho, Nishi-ku,
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The prizewinner shall make a presentation based on his/her work at the Closing Ceremony of the 9th World Congress, Amsterdam on Friday July 3 1998 and the paper shall be duly published in *Prosthetics and Orthotics International*. The President and Executive Board of the International Society for Prosthetics and Orthotics and the Blatchford family reserve the right to withhold the Prize should no suitable candidate be identified.