A modified shank for below knee prosthesis

Mahmud R Ismail 1,2, Yassr Y Kahtan1, and Saif M Abbas 1

1 Prosthetics and Orthotics Engineering Department, Al-Nahrain University, Iraq
2 Author to whom any correspondence should be addressed

Abstract A below knee (BK) prosthesis is a device used to compensate for missing limb segments in patients with BK amputations as an aid for locomotive and other daily activities. It consists of three main parts: socket, shank, and artificial foot. The shank is merely a shaft of a suitable length to serve as a frame structure to transfer biomechanical forces between the socket and foot. In traditional BK prostheses, the shank is made from rigid metallic or composite material tube (Teflon), while in the current work, a modified version is suggested with internal stiffness and damping characteristics to act as energy storage and shock absorber. A mathematical model was thus employed to analyse the impact behaviour from walking gait, and the main parameters studied were the effective mass of impact, loading rate, and knee pre-swing angular velocity. The modified prosthesis was tested with a BK amputation patient and compared with a traditional prosthesis in force platform and treadmill tests. The results showed some improvement in patient gait parameters; at stance phase, it smoothed the GRF curve; at heel strike, both the effective mass of impact and loading rate increased by 16% and 22%, indicating good impact reduction, and at pre-swing, it increased the initial flexion angle and angular velocity by 13% and 15%, respectively, enhancing the swing phase.

1. Introduction

Lower limbs are subject to amputation for many reasons, including disease, accidents, and deformity. Lower extremity amputation may occur at any leg joints, that is, at the hip, knee, or ankle, or between them. Thus, below knee (BK) amputation refers to any amputation on the femoral bone, which will leave residual limb portions, called the stump, of varying sizes.

BK patient lose many joints, muscles, tendons and ligaments vital for performing the required movements for a normal gait; it is thus necessary for any prosthesis to function mechanically to replace such functions. Any improvement in this regard will also produce more comfort for the patient.

Researchers have made many important developments in terms of such devices to improve patient gait cycles. These developments are classified as either passive or active, and for BK prostheses, the main part examined by researchers is the artificial foot, which is intended to simulate normal foot function in terms of absorbing shock and performing various joint articulation functions. Wang Qining et. al. (2010) [1] attempted to design a powered knee ankle prosthesis based on a variable compliance actuator. They suggested a design focused on the transfer of negative energy dissipated by the knee at normal walking pace to the ankle joint at plantar flexion as positive energy, employing a sliding bar with mechanically adjustable compliance and a controllable actuator. This arrangement reduced the total weight of the system and increased autonomy and assistance levels for the patient. Flynn et. al. (2014) [2] attempted a model based on analysing human gait to develop a prosthesis with passive compliant ankles. They used a 2D seven-link model with the addition of hip actuation, knee joints, flat feet, and torsional springs to deliver compliance with ankle joints; two gait events were considered: heel-strike and toe-off. Their results were extended to help understanding of bipedal walking the design of smart ankle foot prostheses and passive robots. Zou
et al. (2015) [3] used a carbon fibre ankle on the foot to improve gait by increasing ankle plantar-flexor power and improving plantar-flexor ankle joint moment as compared with a posterior leaf spring made of thermoplastic materials. Finite element simulation was employed to validate this, with experiments under the same boundary conditions and loading effects. The experimental tests showed that carbon fibre performed better than thermoplastic. Addison and Lieberman (2015) [4] investigated footwear stiffness effects on the impact at foot heel. They presented a model to predict the trade-off between impact loading rate and effective mass, performing several tests on 19 participants wearing three different footwear types to examine walking and running gait. The results showed that the effective mass increased with a decrease in footwear stiffness; the model can thus be used to deduce the effect of stiffness on injury risks during walking and running.

With regard to the shank of prostheses in particular, limited effort has been invested in the improvement of this part except by Lee (2006) [5], who examined a flexible shank made from a single piece of thermoplastic material used with solid ankle cushioned heel to evaluate the gait of unilateral BK amputees. The shank was flexible, with an elliptical cross section, and this was compared with the thicker circular conventional rigid material shank. The more flexible shank was found to reduce the ground reaction force at the beginning and ending of the stance phase.

The aim of the present work is to attempt a new design of shank with added internal stiffness and damping to act as energy storage and shock absorber. The effect of the modified shank on patient gait and impact is thus investigated and compared with the traditional shank.

2. Theory

Figure 1 shows a linkage and lumped model of a BK prosthesis used for investigating the effect of impact.

![Figure 1. Impact model of lower prosthesis (a) linkage model (b) lump model](image)

Many researchers have used the concept of “effective mass” as an index to deduce impact effect. Effective mass is defined as that portion of body mass that decelerates to zero during the impact peak [4]. The ground reaction force (GRF) curve of gait shows two loading peaks during the stance phase of gait: one after the loading rate of heel stride and one near the termination of the toe. Focusing on the impact at the heel the momentum equation can be written as [6]

$$\int_{t_1}^{t_2} (F_G - m_e g) dt = m_e (v_2 - v_1)$$

(1)

where \(t_1, t_2\) are the starting and terminating times of impact; \(v_1, v_2\) are vertical velocities at starting and termination; \(m_e\) is the effective mass; and \(F_G\) is the vertical ground reaction force.
Solving equation (1) for effective mass gives

\[ m_e = \frac{\int_t^t \frac{F_g}{\Delta v + g \Delta t}}{\Delta v + g \Delta t} \]  

(2)

It is important to note that the time and velocity values are evaluated using kinematic analysis of the lower leg in the period between the onset and zenith of heel impact. The other important factor used to compare the effect of impact at the heel is the vertical loading rate, represented by the slope of the line from the initial contact to the impact peak of the GRF curve. This loading rate represents how quickly the impact force is applied, with a steeper slope indicating more rapid collision. The loading rate can be evaluated as

\[ F' = \frac{\Delta F}{\Delta t} \]  

(3)

where \( \Delta F \) is the loading change from initial to peak, assuming a linear relationship at time \( \Delta t \).

3. Experimental Work

3.1: Shank Design and Construction

Figure 2 shows a schematic drawing of the suggested modified shank, which consists of three main parts: an inner tube, an outer tube, and a rubber pad. The rubber pad connects the two pipes by means of two pins and bolts, allowing the shank to be assembled to the proper length depending on patient requirements and allowing flexible movement between the two pipes. The required stiffness value can thus be selected by using a suitable rubber pad to match the patient’s weight and the maximum required deflection. Table 1 presents the main dimensions and materials of the modified shank sample used in testing. The shank was also provided with lock bolt to prevent relative motion between the inner and upper pipes; in the locked position, the shank acts as a rigid (traditional) shank whilst in the unlocked position, the shank is flexible.

![Figure 2: Schematic drawing of the modified shank](image-url)
Table 1: Main dimensions and materials of shank sample

| item          | Diameter (m) | Length (m) | Thickness (m) | Material     |
|---------------|--------------|------------|---------------|--------------|
| Inner tube    | 0.025        | 0.200      | 0.001         | Aluminium    |
| Outer tube    | 0.031        | 0.200      | 0.001         | Aluminium    |
| pad           | 0.020        | 0.030      | -             | Rubber       |
| Inner pin     | 0.003        | 0.025      | -             | Cast iron    |
| Outer pin     | 0.003        | 0.031      | -             | Cast iron    |
| Bolts and nuts| Φ 10         | -          | -             | Stainless steel |

Figure 3 shows the rubber pad used in the design. The stiffness of the rubber pad was measured experimentally using a Testometric Materials Testing Machine, in which the load was applied at 0.5 mm/sec as recommended for rubber materials. The resulting load-deflection line is shown in figure 4. From this figure, the stiffness can be easily determined based on the slope of the line, in this case 198 kN/m.

Figure 4. Load-deflection line for rubber pad

Photographs of the modified shank and its construction are shown in figure 5.
3.2: Case Study
In order to investigate the behaviour of the suggested shank and to compare it with a traditional shank, a BK amputation patient (male, 71 kg in weight and 168 cm tall) offered as a volunteer. The patient has his own BK prosthesis and has experienced good rehabilitation, having used the device for about ten years. For testing purposes, the shank of his prosthesis was replaced with the modified shank, as shown in figure 6.

To explore the kinematic and kinetic behaviours of the shank in the flexible and rigid states, two tests were carried out using treadmill and force platform devices. In the treadmill test, the patient was allowed to walk at 0.78 m/sec (normal walking speed) [7] as shown in figure 7-a, and videos were taken to allow analysis of the leg movements for the full gait cycle. The recorded video was analyse using KINOVIA software which allowed the kinematics of joints and leg segments to be determined and used for impact analysis. In the force platform test, the patient was allowed to walk at normal speed on the plate sensor pad, as shown in figure 7-b. The results of the ground reaction force and gait cycle parameters were then available from the...
software analyser provided. For comparison purposes, both tests were repeated for locked and unlocked shank positions.

(a)                               (b)
Figure 7. (a) Treadmill test and (b) force platform test

4. Results and Discussions
The main gait parameters related to the force platform test are given in table 2.

Table 2 Force platform gait parameters for elastic shank

| Gait Table                        | Elast. Shank |
|-----------------------------------|--------------|
| Number of Strikes                 | 30           |
| Cadence (steps/min)              | 99.1         |
| Gait Time (sec)                   | 12.11        |
| Gait Distance (cm)                | 940.0        |
| Gait Velocity (cm/sec)            | 77.6         |
| Gait Velocity/Leg Length (LL/sec) | n/a          |

Figures 8 and 9 show the vertical ground reaction force of the rigid and flexible shank, respectively. A comparison between the two graphs shows that the trends of these curves become more similar to a normal curve in the flexible configuration, as this contains two peaks of force near the start and termination of the stance phase. The heel stride peak is also larger for the rigid shank.

Figure 8. GRF of the rigid shank
The kinematic data for the vertical velocity of the foot heel and knee were determined by analysing the video of walking gait on the treadmill; open source KENOVEA 0.8.15 software was used in this analysis. For heel impact, five frames prior to impact were selected and “path tracking analysis” was chosen to track the vertical displacement of the foot heel. Figure 10 shows two key image examples used in evaluating impact velocity. The relationships between vertical displacement and the time of the five frames or key images are plotted in figure 11. Impact velocity is thus found by differentiating the displacement curve. In this work, this is done by using the best linear fit, and the slope of this line represents the velocity; the correlation coefficient is $R^2 = 0.88$. The procedure was repeated for the rigid locked and flexible shank cases, and using equation 1, the effective mass can be calculated by means of equation 2. The loading rate was evaluated using the slope of the loading portion from zero to the first loading peak as seen in figures 9 and 10 and equation 3. The main data and results of impact and loading rates for the two cases are shown in table 3.
Table 3. Effective mass and loading rates for rigid and flexible shank

| Case            | Impact period $\Delta t$ (sec) | Speed change $\Delta v$ (m/s) | Effective mass $m_e$ (%BW) | Loading rate $F\,'$ (BW/sec) |
|-----------------|---------------------------------|-------------------------------|---------------------------|-----------------------------|
| Rigid shank     | 18.6                            | 0.52                          | 5.5                       | 35.6                        |
| Flexible shank  | 22.3                            | 0.58                          | 6.4                       | 43.3                        |

Table 3 shows that the effective mass of impact is increased by 16% and the loading rate by 22% for the flexible shank; thus, using the flexible shank tends to spread the impact force over a wider time period as compared with the rigid shank, offering more comfort to the patient at stance phase as the flexible shank acts as a shock absorber.

The other important factor associated with gait kinematics is the angular velocity of the knee, which enhances the pre swing phase. Again, the KINOVEA program was used to calculate the angular velocity of the knee joint for this purpose; the angular displacement of the knee was measured for ten frames just before and just after the toe off event (pre swing), as shown in figure 12. The relationship between the angles and time are plotted in figure 13, with the points are best fitted to a line ($R^2=0.959$) and the absolute angular velocity calculated from the slope of that fitted line. The kinematic data for the knee joint are collected in table 4.

Figure 12. Measuring knee angles using KINOVEA

Figure 13. Angular displacement and time relationship for the knee joint.
Table 4. Kinematic data for knee for rigid and flexible shank

| Case           | Knee flexion | Pre swing angle (deg.) | Knee angular velocity (deg/sec) |
|----------------|--------------|------------------------|-------------------------------|
| Rigid shank    | 44.7         |                        | 240.3                         |
| Flexible shank | 50.5         |                        | 277.1                         |

Table 4 indicates that both knee extension angle and angular velocity of knee are increased for the flexible shank, by 13% and 15%, respectively. Thus, the flexible shank can effectively enhance the pre swing of the prosthesis to help the patient attain similar knee flexion to normal gait, which is about 54 degrees [7]. These increases can be attributed to the ability of the flexible shank to store kinetic energy in the elastic phase in the period between the starting and mid stance phases and to restore this as kinematic energy in the pre swing.

5. Conclusions
Comparison between the kinematics and kinetics results for the modified (flexible) and rigid shank indicates good improvements in the characteristics and behaviours of patient gait when the modified shank is used:

1. The modified shank increases the effective mass of impact and loading rate and acts as shock absorber. The results showed increases of 16% and 22% in terms of effective mass and loading rate, respectively.
2. It enhances knee flexion and angular velocity by 13% and 15%, respectively, at pre swing as compared with the rigid shank.
3. A smoother and closer-to-normal GRF curve is observed when using the modified shank.

6. References
[1] Wang Qining, Huang Yan and Wang Long 2010 Passive dynamic walking with flat feet and ankle compliance. J. Robotica 28 (3).
[2] L. Flynn, J. Geeroms R. Jiménez-fabían and D. Lefeber 2014 Development of a powered knee-ankle prosthesis for transfemoral amputees. Provider: citeseer
[3] Dequan Zou, Tao He, Michael Dailey and Kirk E. Smith 2015 Experimental and computational analysis of composite ankle-foot orthosis. J. Rehabilitation Research and Development 51 (10)
[4] Brian J.Addison and, Daniel E. Lieberman 2015 Tradeoffs between impact loading rate, vertical impulse and effective mass for walkers and heel strike runners wearing footwear of varying stiffness. J. Biomechanics 48 p.1318.
[5] Winson C. C. Lee Ming Zhang, Peggy P. Y. Chan, and David A. Boone. 2006 Gait Analysis of Low-Cost Flexible-Shank Transtibial Prostheses. IEEE transactions on neural systems and rehabilitation engineering 14. (3).
[6] Brian James Addison 2015 The biomechanics and evolution of impact resistance in human walking and running, PhD thesis, Harvard University, Cambridge, Massachusetts,
[7] Benjamin F. Mentiplaya,, Megan Banky, Ross A. Clark, Michelle B. Kahna,,Gavin Williams 2018 Lower limb angular velocity during walking at various speeds, J. Gait & Posture. 65, p190.

Acknowledgments
The authors wish to acknowledge the assistance from technical staff at the Prosthetics and Orthotics Engineering Department, College of Engineering, Al-Nahrain University. Great thanks also go to Mr. Karar Ali for his participation in testing the prosthesis.