Design and Analysis of a Novel Artificial Ankle-Foot Joint Mechanism

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Abstract. Globally, more than one million limbs are amputated annually. Theoretical and experimental studies thus been undertaken in many countries to develop prosthetics to help alleviate the suffering of people with amputated limbs. This study aimed to design and simulate an artificial ankle joint to mimic human walking on sloped surfaces efficiently. The new design also attempts to mimic the shape, the size and the range of angle motion of the human ankle joint while simultaneously reducing the cost and weight of the prototype prosthetic ankle unit and increasing durability and responsiveness. The weight of the individual user is the main factor controlling the engagement and disengagement of the artificial ankle joint during the gait cycle, and the new design for the artificial ankle joint was prepared and tested using CATIA and ANSYS software. The suggested model was shown to be effective and reliable and able to adapt to multiple different surfaces. It also achieved high flexibility under various weights of users. Key words: Design, Artificial, Ankle Foot, Joint Mechanism.

1. Introduction

The number of amputees globally is increasing due to the effects of war, disease, and traffic accidents. Many accidents affect the legs below the knee, and research work in this area thus often focuses on below-knee prostheses. In Iraq, rehabilitation centres offering prostheses and orthotic devices use older techniques in manufacturing these. Recently, however, Iraqi researchers have focused on designing and manufacturing below and above knee prostheses to support the ongoing on powered prostheses [1,2,3,4,5].

All types of ankle-foot prostheses generated thus far have lacked the high efficiency that must be provided to support optimum function [6]. While distinctive various keel designs that provide energy-storage for walking and running are available their gap in the ankle joint is unattractive. Further, both level and ramped surfaces impose the necessity for human gait to adapt to create suitable patterns to maintain a body’s equilibrium [7-10]. The relationship between the ankle alignment and a ramped surface is in line with the net equilibrium of the body [7], and changes in the orientation of the trunk and pelvis are required for adaption to walking between level and ramped surfaces [8]. Changing the slope of the walking surface also alters both the level and orientation of the contact surface [11]. Andrew et al.
noted a non-significant effect on the roll-over shapes of the ankle-foot and knee-ankle-foot systems based on symmetric increases in the weight about the trunk during slow and normal walking that were altered somewhat on running [12]. A newly developed portable real-time stress monitor was then used to evaluate the internal stresses in the soft tissues under the tibia of the residuum of 18 disabled people [13], as previous research had indicated that the non-disabled ankle-foot system adapts to a walking surfaces without collecting any data from the previous steps [14].

The main objectives of the current work were to design a new ankle-foot joint mechanism that can accommodate changes in terrain perfectly. The design was intended to offer a weight-activated locking/unlocking system that operates without conscious effort from used and which could absorb the reaction of the ground forces, as well as reducing the ankle-foot joint mass. The new design thus took into consideration patients’ physical and psychological attributes and age groups, as well as their financial needs.

2. Suggested Model

A short survey was undertaken to achieve a design suitable for a large number of potential users. Specific data was collected based on the measurement of five individuals without physical impairments as shown in Table 1. This data includes the circumference of the ankle joint, the weight of the individual, their foot length, and their age. The circumference of the SPF (the largest circumference) of the design (Figure 2 - A) was thus based on the average for the assessed individuals.

| No | Age (yrs.) | Weight (kg) | Foot length (cm) | Circumference (cm) |
|----|------------|-------------|------------------|-------------------|
| 1  | 20         | 55          | 25               | 7.3               |
| 2  | 31         | 60          | 26               | 7.6               |
| 3  | 29         | 65          | 27               | 8.2               |
| 4  | 33         | 70          | 26.5             | 8.5               |
| 5  | 24         | 75          | 28               | 8.4               |

The model of the artificial ankle joint, as shown in Figure 1, was designed using the data in Table 1 in CATIA V5R20 software. The model was then exported to the ANSYS workbench for Transient Structural Analysis. The material for the stiff parts of the artificial ankle joint was set as structural steel (Table 2), and after exportation, the elements of the joint were meshed and subjected to five different loads in the same manner as loading a normal ankle joint under the same conditions. The weights applied were 55kg, 60 kg, 65 kg, 70 kg, and 75 kg.

![Figure 1. Cross section through the artificial ankle joint model](image-url)
Table 2 The properties of the steel used in the model design

| Name of the part | SPF, lock/ unlock pins, spherical joint, beams, SPF based, adapter based, and keel ring | Foot plate | Compression spring |
|------------------|---------------------------------------------------------------|------------|------------------|
| Material used    | Structural steel                                             | plastic plate | Structural steel |
| Density (Kg m^-3) | 7850                                                          | 7850       |                  |
| Young’s modulus (pa) | 2E+11                                                        | 2E+11      |                  |
| Poisson’s ratio  | 0.3                                                           | 0.3        |                  |

The artificial ankle joint functions were entirely associated with the application of weight, as shown in Figures 1 and 2. When the heel of the foot contacts the floor (heel contact), the weight of the user is applied increasingly to the ankle joint. As a result, the compression spring is displaced by its maximum vertical displacement (ΔX = 4.5mm) due to the effect of the weight. The artificial ankle joint absorbs the energy during this period within a spring, until the lock/unlock pins are disengaged as the deformation of the compression spring enables the adaptor to be dorsiflexed or plantar flexed. The pins are fixed to the Spherical Plate Frame (SPF) segments by means of screws. The SPF (figure. 2, A) begins to bear the incremental weight after the adaptor-based (AB) segment is fully compressed (Figure 2 - A); the AB thus sits over the SPF part during the stance. The SPF is deformed by a specific vertical distance under the effect of the person’s weight as a means of absorbing the reaction of the ground forces. The energy is thus stored both in the spring and the SPF.

The artificial ankle joint consists of several components that work together to produce two principal movements, dorsiflexion and plantarflexion, which give abduction and adduction movement. Additionally, the axial deformation of the compression spring and the deformation of the SPF creates a vertical displacement that absorbs the reactions of the floor. This work focuses on only two parts of this movement: the dorsiflexion and plantarflexion movement of the AB segment and the deformation of the spring and SPF. The AB segment plays an essential role in controlling the engagement and disengagement of the artificial ankle joint’s movements. Four lock/unlock pins are used, fixed to the AB segment, where two of them prevent the movement of the AB section during the swing phase and the other two pins engage with the SPF segment when the heel makes contact with the floor (HC) and in the flat foot phase (FF).
Figure 2. Artificial ankle joint components with virtual socket and foot.

A compression spring was used to control the down-up displacement of the AB section and to absorb the ground reactions. The material chosen for the compression spring was a structural steel, and the index of the spring was 5.5. The upper side of the spring was attached to the lower surface of the AB segment, whereas the lower side of the compression spring was attached to the spherical joint. As a result, the spring damping occurred between the AB segment and the spherical joint.

The SPF segment is the most important element in this artificial ankle joint design; this has four legs are spherically curved, with the posterior and anterior legs fixed on a beam jointed to the SPF segment, while the other two legs (lateral and medial) allow a small pivot movement to the lower parts of the artificial ankle joint (SPF base, keel ring, and footplate) with regard to the fixed beam. This pivoting movement allows the foot to manage an uneven floor proficiently. The SPF segment also supports the spherical joint parts. The two main functions of the SPF are to absorb the impact reaction forces of the ground and to adjust the kinematics and kinetics of the moving parts. A bumper was placed between the SPF and AB sections to return the AB segment to its original position after a step. Two beams were used to fix the spherical joints to the SPF segment and the SPF to the lower sections.

Three levels of lock/unlock steps were created for the SPF segment, positioned specifically on the two sides of the AB section (Figure 2). The unlock steps create an obstacle to the lock/unlock pins during an ankle-foot roll-over as well as allowing the joint to adjust to the required angle range of the plantarflexion voluntarily (10.2°, 20.2°, and 31.2°, see Figure 3). The lock/unlock pins are released after the push off period and the disengagement of the lock/unlock pins on the AB section is achieved after the load is released. As a result, the potential energy stored in the spring and SPF segment is released to return the ankle joint elements to their original shape during the swing phase. The bumper then repositions the AB section and the other elements that it is connected to (Figure 3).

The arc shape of the keel is designed to mimic the ankle-foot rocker shape during normal walking. This design allows the foot plate to bend until it adapts to the keel rocker shape [13]. The artificial ankle joint can manage defluxion of just over 20°, as shown in Figure 3.
Figure 3. Angle range of plantarflexion.

The virtual socket and the other connecting parts in plantar flexion to 10.2° when the lock pins engage with the lock steps (a), lock pins and lock steps engaged at a 20.2° plantarflexion angle (b), and lock pins and lock steps engaged at a 31.2° plantarflexion angle (c).

Figure 4. Angle range of plantarflexion

The virtual socket and the AB section plantar flexed with a maximum angle of 19.5° (a), the lock pins and lock steps engaged at a 31.2° plantarflexion angle (b).

3. Results and Discussion

3.1 Result

An analytical model of the compression spring deformation was established using ANSYS workbench (transient structure). The spring material was set as structural steel and the stiffness behaviour of the spring was set to be flexible. The resulting deformations of the compression spring along the Z global coordinates wee as shown in Figure 5. The deformation of the spring was 4.5 mm as a displacement under the effective weight of the user. Figure 5 shows the variation of maximum directional displacement of the spring, an indication of the spring stiffness. Prior to manufacturing such a spring, the variation in spring behaviour should be defined specifically with regard to the human gait cycle, and this analysis meets this requirement. The directional deflection of the spring varies during the heel strike.
for the first four seconds then becomes constant due to contact with the SPF. Using this information, an effective spring design can be achieved. The AB part was also displaced by the same displacement as the spring along the same coordinate. The AB segment was disengaged under this deformation (4.5 mm), allowing the AB section to dorsiflex and plantar-flex at specific angles (Figure 5-A). The results from the ANSYS workbench simulation indicated that, after the total displacement of the spring (4.5 mm), excess weight was carried by the SPF element and the spherical joint element.

The compression spring was compressed at the display by a vertical distance along the z global coordinate of 4.5 mm (Figure 6-A). The load was transferred to the SPF after this deformation. The ANSYS workbench window shows the maximum directional deformation of the compression spring as -4.5169 mm and the minimum directional deformation of the compression spring as 0.04790 mm (Figure 6-B).

After that, the AB segment begins the plantar flexion period, as shown in Figure 6-A, based on the disengagement of the lock/unlock pins. The maximum angle achieved was 32°. The AB segment then returns to its original position while the lock pin engages with the lock steps on the SPF section (Figure 6-B). The AB part is then dorsiflexed about the ankle axis joint, from 0 to 22°. These measurements are in agreement with the ISO 10328 standard, allowing amputees to use this mechanism with the required degrees of freedom plantarflexion and dorsiflexion with reasonable weights.

Figure 5. Spring behaviour under the direct effect of user weight.

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Figure 6. The range of motion of the artificial ankle joint

A) Range of dorsiflexion and plantar flexion about the axis rotation of the ankle (-32° to 22°),
B) Maximum plantarflexion (-32°),
C) Maximum dorsiflexion (22°), and
D) AB part returned to its original position.

Any excessive weight applied is taken up by the SPF section (Figure 7). The weight of the individual was applied gradually, using tabular data, in ANSYS workbench. This action was repeated for individuals of 55 kg, 60 kg, 65 kg, 70 kg, and 75 kg. The results showed that the total deformation of the SPF segment along the Z axis was 0.2 mm, with the SPF part remaining stiff under the maximum weight (75 kg) in a manner acceptable under ISO 10328 prostheses standards. Figure 7 shows the maximum total deformation over time, with variation in the first four seconds of the gait cycle as discussed earlier.

Figure 7. Total deformation of the SPF section under the effective weights of the individuals assessed
The vertical axis shows the bending of the material in the SPF section when the user is standing directly over the artificial ankle joint (Figure 7-A). Here, the maximum deformation of the SPF is equal to 0.15209 mm and the minimum deformation is equal to -0.043127 mm.

3.2 Discussion

In order to discuss the new artificial ankle joint, the designed mechanism was analysed by using the software to examine variation with different types of floor level, assessing the joint’s capacity to adjust to different environments. Based on the results, the following points emerged:

1. Several positive indicators that the joint could adapt to different slopes emerged; this depended on the applied loading and the type of foot movement. For instance, the angle ankle joint maintained the same path even where the user weight was applied to follow the level of the floors.

2. The flexibility of the model means that all parts return to a consistent neutral position after each gait cycle step, allowing angle joint alignment to occur independently for every step of walking. These characteristics are in contrast to those of a standard prosthesis that, although highly dynamic, shows widely dissimilar ankle angle trajectories when walking on surfaces of different slopes as well as overlapping ankle torque-angle relationships, behaviour characteristic of a device with only has a single set-point and a direct relationship between ankle angle and ankle torque.

3. The artificial ankle joint was able to achieve late-stance plantarflexion (10°, 20°, and 32°), unlike a standard prosthesis. Additionally, even with the maximum weight of user, the vertical height of the spherical joint axis was displaced by only 0.2 mm about the original position. This should ensure stability of position in the ankle joints axis during postural standing.

4. The structure of the joint held up well to the effective weight and the analysis showed that the structure was more rigid and durable than prior versions. Moreover, the structure of the model
responded to the effect of the weight, with the joint storing the energy of the user applying a load on the joint and releasing this on it unloading. The mass of the joint was thus minimised compared with previous versions [6], with the total weight of the model reduced to 0.55 kg, based on new elements designed to enhance the functionality and durability of the model. Furthermore, the external shape of the joint mimics the shape of natural joints.

5. The engagement and disengagement mechanism for the artificial ankle joint was voluntary and thus easier to use than previous mechanisms. The number of parts controlling the joint movements was limited, and the external profile ruggedly formed to enhance durability. The external shape of the artificial joint was optimised to mimic the natural shape of the ankle joint. As the model is designed to work based on the weight imposed on the joint, the user does not need to control the engagement/disengagement of the joint during walking, enhancing concentration.

6. There are several limitations to this work. The design assumes that the footplate is a simple plastic plate, and the study thus did not assess the energy that could be stored and released through the gait cycle. In addition, this work assumed that the neutralising bumper was sufficient to reposition the AB segment, and the necessary interactions were not included during the simulation of the model. In the future, it may thus prove necessary to replace the neutralising bumper with a compression spring to optimise the joint functions.

4. Conclusions
The model analysis demonstrated strong evidence of efficient interaction between the joint and the weight of the user. It also showed that the joint stored energy in the spring and SPF segment during the stance phase and released it during the swing phase. The modelling of this new artificial ankle joint demonstrated good adaptation to various types of surfaces, and further work should focus on refining the design, simulating the ankle-foot behaviours on inclines, declines, and eversion and inversion surfaces, prior to manufacturing the system and testing its durability.

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