INTERNATIONAL JOURNAL OF INNOVATIVE TECHNOLOGY & CREATIVE ENGINEERING (ISSN:2045-8711)
VOL.1 NO.8 AUGUST 2011

Toward a 21st Century Crutch Design for Assisting Natural Gait

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***Abstract*—In order to resolve the disadvantages of conventional crutch designs (i.e. underarm and forearm), a novel crutch design is presented with the name of *21st Century Crutch*. Prior the design, human natural treadmill walking is monitored by a 3D Motion Capture System and acquired a reference end–foot trajectory with a '*teardrop shape*'. Considering the design objectives, natural human walking and comfort, and other factors such as load capacity and weight of the device, the final design was determined. In order to satisfy the design objectives, a kinematic synthesis previously worked by Robson and McCarthy [14] is applied to test if the end–foot trajectory of designed crutch smoothly follows the desired reference, '*teardrop shape*' in the vicinity of two specified task points, heel strike and toe off. For that reason the leg was synthesized and animated in Mathematica as a RR planar kinematic chain, where the first hinge joint/fixed pivot was located at the hip and the other hinge joint/moving pivot is at the knee joint. A prototype of the final design was fabricated and its performance was tested by 2 mph treadmill walking.**

Keywords: Crutch design, Human walking, Planar chain

I. INTRODUCTION

The crutch is the simplest and reliable way to compensate the mobility of people with lower limb injuries by supporting their body weight during locomotion in daily life (e.g. ascending/descending stairs and walking). It provides a stable environment for recovery by allowing the injured body part in a load free condition. It is known that the crutches have been used for 5,000 years [15]. People used fallen tree branches as supporting sticks to help balancing or ambulating wounded body. From its primitive forms, the current configurations of underarm and forearm crutches have been evolved through various empirical designs. The first US patent for a crutch was issued to Tuttle (US patent No. 332,684) [16]. The first commercialized form of forearm crutch design was patented by a French mechanical engineer, Schlick, as a *walking stick* in 1917 (US patent No. 1244249) [16]. From these patented designs, numerous modification works have been patented for comfort and safety.

Though the current crutch designs are inexpensive solutions to fulfil their main function, body weight support, they have non–negligible disadvantages follows:

1) since the armpit or forearm are not supposed to support a large load, normally, continuous stress on them can cause degraded motion control and even nerve damages [2],
2) important upper limb functional motions in daily living (e.g. arm reaching motion, hand grasping and manipulating) are limited,
3) decreased degrees of freedom (DOF) can induce compensatory joint motions and unnatural locomotion, which can be possibly developed as chronic pathologies after long term of usage, and
4) ambulation with underarm and forearm crutches demands high cost of energy consumption [17].

The iWalk–Free (iWALKFree, Inc., USA) [18] shows a typical example of novel crutch design to overcome the disadvantages described above. With a stable load support through 90° flexed knee, the device could reduce discomfort significantly by freeing upper limb motions. However, it is only applicable to injuries on below tibia and fibula region (i.e. foot and ankle). Also, due to lack of knee joint DOF, the hip and pelvic joints tend to make an abnormal motion pattern to ensure the foot clearance during the swing phase of gait.

The idea behind the proposed crutch design is at assisting ambulation of patients who have injured one of their lower legs (i.e. below femur, including knee joint). These injuries include ankle sprain, fracturend/or cracks on tibia, fibula, foot bones, or any combination of them. The skeletal anatomy of human leg is shown in Fig. 1.

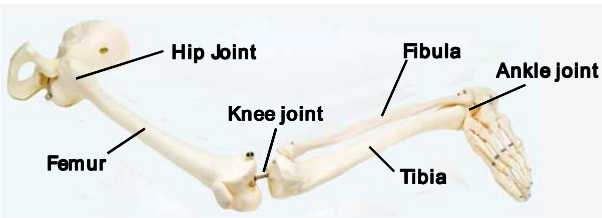


Fig. 1. Skeletal anatomy of human leg (image source [5])

In this paper, a novel crutch design is developed with the name of *21st Century Crutch*. In order to cope with significant disadvantages of conventional crutch designs, two distinct design objectives, natural walking motion and comfort, are set. On its way to the final design, a preliminary concept was developed and tested for comfort and natural gait trajectory of the foot. Next, after the preliminary concept, a planar chain with RR (Revolute– Revolute) joint was tested in Mathematica. A kinematic synthesis of linkage design with acceleration specification is adopted to achieve the design objectives (i.e. natural walking motion and comfort). The goal of this synthetic approach is to obtain the solutions to a given task specification (i.e. end–foot trajectory of normal treadmill walking) in order to test the design of the mechanical linkage that can move the end–foot smoothly through the specified task.

Unlike syntheses with position and velocity specifications, research in the synthesis of serial chains to achieve acceleration requirements is limited. It is primarily found in the synthesis theory for planar RR chains and the work by Chen and Roth [10] for spatial chains. The use of second order effects first appears in the analysis of grasping in a work by Hanafusa and Asada [11], where planar objects are grasped with three elastic rods.

Rimon and Burdick [12,13] showed that acceleration properties of movement can be used to effectively constrain a rigid body for part–fixing and grasping applications. Robson and McCarthy [14] presented a technique for deriving geometric constraints on position, velocity and acceleration from contact and curvature task requirements. These constraints yield design equations that can be solved to determine the dimensions of the serial chain. In this paper, despite the fact that the dimensions of the serial chain are pre–determined (i.e. size of the crutch must be customized to the human subject), the developed kinematic synthesis theory is applied to test the end–foot trajectory and its smoothness.

The detailed design objectives and required specifications are described in Section II. After that, the Section III considers the synthesis of planar chains to guide the foot through multiple number of separate positions [8,9]. The required kinematic specifications in the synthesis are number of task positions with specified end–foot velocities and accelerations. The Section IV introduces the preliminary design and is followed by the Section V which describes the final design and its prototype fabrication. The experimental result is shown in the Section VI. Finally conclusion and future works are discussed in the Section VII.

II. DESIGN OBJECTIVES AND SPECIFICATIONS

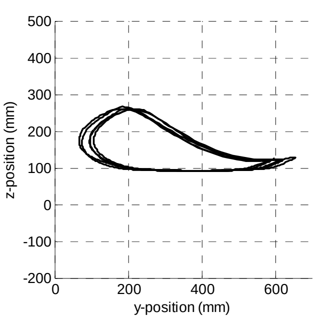
Before the actual design work, the details of design objectives (i.e. natural walking motion and comfort) and other factors are described in this section.

A. *Natural Walking Motion*

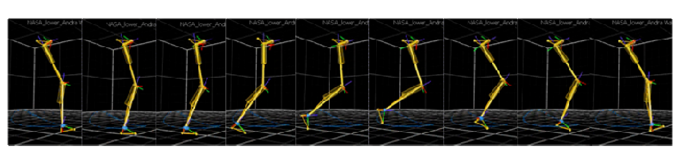
Like iWalk–Free design, the load supporting structure should be transferred from the upper limbs (e.g. armpit or forearm) to the lower limbs. This allows the user to keep more natural walking motion compare to conventional crutch designs (see the disadvantages of conventional crutch design described in Section I). For realizing even more natural walking, the proposed novel crutch in this paper is designed to mimic the normal end–foot trajectory so that important gait events and properties (e.g. heel strike, toe off and foot clearance) can be performed smoothly and effectively.

In order to acquire a reference gait data, a person's normal treadmill walking with 2 mph speed was collected via 3D Motion Capture System (Vicon MX, Vicon Inc., UK). Eight reflective markers were attached on the subject's right leg as shown in Fig. 2(a) with highlighted circles. Fig. 2(c) represents the single gait cycle obtained by the 3D Motion Capture System. For visualizing the actual end–foot trajectory, the positions of ankle marker on the sagittal plane is plotted (see Fig. 2(b)). Note that the geometrical shape of the reference trajectory looks like a '*teardrop*' for each gait cycle.

(a) Attached marker positions



(b) Acquired foot trajectory



(c) A cycle of walking motion in the 3D Motion Capture System

Fig. 2. Capturing a normal treadmill walking cycle

B. Comfort

Comfort is an important factor which directly affects the subject's ADL (activities in daily living) performance. It is deeply related with the ergonomic design, which pursues to reduce the user's fatigue. In this paper, the secure attachment mechanism and its location on the subject's body are considered to maximize the comfort.

C. Others

1) Load Bearing Capacity

Since the injured patients solely depend on the crutch to support their body weight during locomotion, it becomes the most critical design factor which is directly related to the safety issue.

The crutch should be able to assist most of locomotion in daily living such as level walking, sitting down, standing up and ascending/descending stairs. In order to specify the load bearing capacity, the empirical data on maximum loading of the knee joint during ADL is adopted. According to the study of Kutzner *et al*. [1], average peak resultant forces on the knee joint was 261% BW (body weight) for level walking while the maximum loading was applied during descending stair (346% BW). Therefore, for instance, the crutch should be designed to sustain at least 692 lb to support a 200 lb person without counting a safety factor.

2) Light Weight

As the proposed design is intended to be attached on the lower limb, it imposes an additional mass on the subject's body and alters weight of the limb which induces changed dynamics during locomotion. Also the varied inertia can cause compensatory motions of other DOF (e.g. pelvic and hip joint) during a leg swing motion and they can be further developed as pathological abnormal joint patterns. Therefore in order to minimize these side effects, the device is required to be designed in light weight as possible. As a reference, according to an anthropometric rule, weight of a leg is approximately 16.1% of BW [19].

3) Passive Actuation

The design can contain any moving mechanisms to enhance its functionalities (e.g. less energy consumption, impact force absorption and increased dynamic stability). Higuchi *et al*. [3] adopted linear actuators with telescopic links to reduce user's efforts while using a pair of underarm crutches. Mori *et al*. [4] added a mobile platform which can interact with a sensor integrated underarm crutches to enhance the mobility. However, mechanisms with active actuation systems incorporate complicated control and monitoring algorithms which can possibly cause unintended malfunction. Therefore, for the safety and simplicities in design, a passive actuation mechanism is considered in this paper.

4) High Mobility

In order to maximize the performance on ADL of the proposed design, a high mobility is needed. The high mobility will allow the subject to be able to explore challenging environments (e.g. uneven terrain and obstacles) even after the injury.

5) Usability

Due to the crutch's high frequency of use, its usability should be considered in the design stage. Easy put on and off mechanism and adjustable size mechanism can be typical examples of this factor.

III. KINEMATIC SYNTHESIS OF PLANAR LINKAGES

The kinematic synthesis of planar linkages is adopted to test the final crutch design (i.e. RR chain, see Section V). The pattern of one gait cycle consists of swing and stance phases. Since the leg dynamics of each phase is totally different, two phase transition events, heel strike (i.e. when the foot initiate its contact with the ground at the beginning of a stance phase) and toe off (i.e. when the foot terminates the ground contact at the ending of a stance phase), contain useful information to characterize the dynamics and kinematics of gait.

In this paper, the kinematic specifications (i.e. position, velocity and acceleration) at those two events in the reference gait trajectory will be derived by a planar synthesis proposed in [14]. The derived kinematic specifications are utilized to manipulate the end–foot trajectory of the designed crutch to mimic the desired one (i.e. natural human walking). Also, by applying the derived kinematic specifications to the geometric design equations of RR planar chain, an appropriate position of RR chain's base frame (i.e. attachment location of the crutch on the subject's thigh) can be determined. The purpose of this section in the context of design objectives can be described as follows:

1) obtaining required kinematic specifications at heel strike and toe off events to realize a natural gait pattern (i.e. '*teardrop shape*', see Fig. 2(b)) of the designed crutch, and
2) obtaining an accurate attachment location of the designed crutch on the subject's thigh to enhance its fit, dynamic stability and comfort.

In what follows, a planar synthesis approach recently developed by Robson and McCarthy [14] is presented.

A. Geometric Design of Planar Mechanical Linkages with Task Acceleration Specifications

The walking motion can be assumed as a planar task on the sagittal plane which consists of positioning of the foot, at the point, M^j (j=1,…, n), located on the reference trajectory. If the kinematic specifications at the start (i.e. heel strike) and end points (i.e. toe off) are acquired, it is possible to manipulate the designed RR chain (i.e. crutch) to mimic the reference trajectory pattern. The required kinematic specifications are derived from contact and curvature constraints between the foot and the ground in the specified event positions. The derivation of these constraints is discussed in details in [14]. The movement of foot can be described by the parameterized set of 3x3 homogeneous transform matrix

$$\left[\mathbf{T}(t)\right]=\left[\mathbf{R}(t),\mathbf{d}(t)\right] \tag{1}$$

where **R**(t) and **d**(t) represent a rotation matrix and a translation vector, respectively. A point **p** fixed in themoving body traces a trajectory **P**(t) in a fixed coordinate frame F such that

$$\begin{bmatrix} P_x(t) \\ P_y(t) \\ 1 \end{bmatrix}=\begin{bmatrix} \cos\phi(t) & -\sin\phi(t) & d_x(t) \\ \sin\phi(t) & \cos\phi(t) & d_y(t) \\ 0 & 0 & 1 \end{bmatrix}\begin{bmatrix} p_x \\ p_y \\ 1 \end{bmatrix} \tag{2}$$

or

$$\mathbf{P}(t)=\left[\mathbf{T}(t)\right]\mathbf{p} \tag{3}$$

The next goal is to determine the movement of the foot, as defined by [**T**(t)]. The movement of the foot, relative to a world frame in the vicinity of a reference position, defined by t=0 can be expressed by the Taylor series expansion,

$$\left[T^j(t)\right]=\left[T_0^j\right]+\left[T_1^j\right]t+\frac{1}{2}\left[T_2^j\right]t^2+\dots \quad (j=1,\dots,n) \tag{4}$$

where

$$\left[T_i^j\right]=\left.\frac{d^i\left[T^j\right]}{dt^i}\right|_{t=0} \tag{5}$$

The matrices $[T_0{}^j]$, $[T_1{}^j]$ and $[T_2{}^j]$ are defined by the position, velocity and acceleration of the foot in the vicinity of each task position M^j, respectively. Therefore, a point **p** in M has the trajectory **P**(t) defined by the equation

$$\mathbf{P}^j(t)=\left[T^j(t)\right]\mathbf{p}=\left[T_0^j+T_1^jt+\frac{1}{2}T_2^jt^2+\dots\right]\mathbf{p} \tag{6}$$

Let $\mathbf{p}=[T_0{}^j]^{-1}\mathbf{P}^j$, which yields

$$\begin{aligned}\mathbf{P}^j(t)&=\left[T_0^j+T_1^jt+\frac{1}{2}T_2^jt^2+\dots\right]\left[T_0^j\right]^{-1}\mathbf{P}^j\\&=\left[I+\Omega^jt+\frac{1}{2}\Lambda^jt^2+\dots\right]\mathbf{P}^j\end{aligned} \tag{7}$$

where

$$\left[\Omega^j\right]=\begin{bmatrix} 0 & -\phi_1 & d_{x1}+d_{y0}\phi_1 \\ \phi_1 & 0 & d_{y1}-d_{x0}\phi_1 \\ 0 & 0 & 0 \end{bmatrix} \tag{8}$$

$$\left[\Lambda^j\right]=\begin{bmatrix} -\phi_1^2 & -\phi_2 & d_{x2}+d_{x0}\phi_1^2+d_{y0}\phi_2 \\ \phi_2 & -\phi_1^2 & d_{y2}+d_{y0}\phi_1^2-d_{x0}\phi_2 \\ 0 & 0 & 0 \end{bmatrix} \tag{9}$$

$[\Omega^j]$ and $[\Lambda^j]$ are the planar velocity planar acceleration matrices, respectively, which are defined by the foot kinematic specifications in the vicinity of the task positions M^j.

B. Design Equations

In this paper, the end–foot kinematics is presented as a planar RR mechanical linkage. The design parameters for the planar RR chain are the coordinates **B**=(B_x, B_y) of the fixed pivot, located at the hip joint and the coordinates $\mathbf{P}^1$=(P_x, P_y) of the moving pivot, located at the knee joint, when the floating link is in its start position. The ankle is assumed to be rigid since the final design does not include an ankle joint (see Section V). In each task position, the moving pivot $\mathbf{P}^j$ is constrained to lie at the distance *R* (i.e. length of the moving link, which connects the hip and knee joints) from **B**, which yields,

$$\left(\mathbf{P}(t)-\mathbf{B}\right)\cdot\left(\mathbf{P}(t)-\mathbf{B}\right)=R^2 \tag{10}$$

Note that the link length, *R*, and the length of tool frame, H, which has its tip located at the ankle are known values.The first and second derivatives of (10) provide the velocity constraint equation

$$\frac{d}{dt}\mathbf{P}\cdot\left(\mathbf{P}-\mathbf{B}\right)=0 \tag{11}$$

and the acceleration constraint equation

$$\frac{d^2}{d^2t}\mathbf{P}\cdot(\mathbf{P}-\mathbf{B})+\left(\frac{d}{dt}\mathbf{P}\right)\cdot\left(\frac{d}{dt}\mathbf{P}\right)=0 \quad (12)$$

In order to determine the five design parameters, five design equations are required. Choosing one of the task positions to be the first and applying the relative displacement matrices

$$\left[\mathbf{D}_{1j}\right]=\left[T_0^j\right]\left[T_0^1\right]^{-1} \quad (13)$$

allows the coordinates $\mathbf{P}^j$ to be defined by the moving pivot as follow

$$\mathbf{P}^j=\left[D_{1j}\right]\mathbf{P}^1 \quad (14)$$

By substituting (14) into (10), the constraint equation becomes the position design equation below.

$$\left(\left[D_{1j}\right]\mathbf{P}^1-\mathbf{B}\right)\cdot\left(\left[D_{1j}\right]\mathbf{P}^1-\mathbf{B}\right)=R^2 \quad (i=1,\ldots,n) \quad (15)$$

In the crutch design, n=2. Notice that $[D_{11}]$ is a 3 x 3 identity matrix. From the previously defined 3 x 3 velocity matrix, (8), the below can be derived.

$$\frac{d}{dt}\mathbf{P}^j=\left[\Omega^j\right]\left[D_{1j}\right]\mathbf{P}^1 \quad (16)$$

By substituting (14) and (16) into (11), the velocity design equations for the two specified positions can be represented.

$$\left(\left[\Omega^j\right]\left[D_{1j}\right]\mathbf{P}^1\right)\cdot\left(\left[D_{1j}\right]\mathbf{P}^1-\mathbf{B}\right)=0 \quad (j=1,\ldots n) \quad (17)$$

Similar to (16), the below can be derived from the defined 3 x 3 acceleration matrix, (9).

$$\frac{d^2}{dt^2}\mathbf{P}^j=\left[\Lambda^j\right]\left[D_{1j}\right]\mathbf{P}^1 \quad (18)$$

Also the substitution of (14) and (18) into (12), the acceleration design equations for the start position are derived.

$$\left(\left[\Lambda^j\right]\left[D_{1j}\right]\mathbf{P}^1\right)\cdot\left(\left[D_{1j}\right]\mathbf{P}^1-\mathbf{B}\right)+\left(\left[\Omega^j\right]\left[D_{1j}\right]\mathbf{P}^1\right)\cdot\left(\left[\Omega^j\right]\left[D_{1j}\right]\mathbf{P}^1\right)=0 \quad (19)$$

Therefore, for each of the n task positions, the position, velocity and acceleration design equations become

$$\begin{aligned}
&P_j:\left(\left[D_{1j}\right]\mathbf{P}^1-\mathbf{B}\right)\cdot\left(\left[D_{1j}\right]\mathbf{P}^1-\mathbf{B}\right)=R^2,\\
&V_j:\left(\left[\Omega^j\right]\left[D_{1j}\right]\mathbf{P}^1\right)\cdot\left(\left[D_{1j}\right]\mathbf{P}^1-\mathbf{B}\right)=0,\\
&A_j:\left(\left[\Lambda^j\right]\left[D_{1j}\right]\mathbf{P}^1\right)\cdot\left(\left[D_{1j}\right]\mathbf{P}^1-\mathbf{B}\right)\\
&\quad+\left(\left[\Omega^j\right]\left[D_{1j}\right]\mathbf{P}^1\right)\cdot\left(\left[\Omega^j\right]\left[D_{1j}\right]\mathbf{P}^1\right)=0 \quad (j=1,\ldots,n)
\end{aligned} \quad (20)$$

The algebraic solution to the set of four bilinear equations for an RR chain is presented in [20] and can be applied without any changes for the case of five position synthesis (i.e. design equations (20)).

IV. PRELIMINARY DESIGN AND RESULT

Based on the design objectives and specifications (see Section II), a closed–loop linkage type crutch, TAMC1, is developed as a preliminary design. The main purpose of TAMC1 is to test the performance based on comfort and natural gait.

A. Design

The design of TAMC1 is shown in Fig. 3. In order to protect the injured body part (i.e. lower leg region including knee joint), it is assumed that the user is in a cast, which covers the leg from the foot to just above the knee joint, with a slightly flexed knee (see Fig. 3), or is missing a part of or the entire one's lower leg.

The user's body weight is supported through a seat structure located within the thigh region (see Fig. 3(a)). The large and flat thigh seat allows the body weight to be evenly distributed over the entire contact area. At the same time, not to interrupt the natural hip joint motion, the size and location of seat is carefully selected through experimental trials.

In a gait cycle, stance leg supports the whole body weight by keeping the knee joint nearly full extension state. Therefore the dynamics of entire body during the stance phase can be assumed as a rigid link inverted pendulum. Though there is a slight knee flexion during a stance phase, its range is small and its purpose is on diminishing vertical fluctuation of COM (center of mass) [6]. Adamczyk *et al*. [7] focused on the inverted pendulum like dynamics of the stance leg and showed that the metabolic cost can be reduced by adopting a rolling foot to take advantage of the stance leg dynamics. The TAMC1 design consists of two closed–loop linkages, 5–bar linkage for the upper link (see Fig. 3(b)) and 4–bar linkage for the lower link (see Fig. 3(c)), respectively. The upper link motion mimics the inverted pendulum motion while the lower link emulates the rolling foot motion. In order to reduce any impact forces and to propel the body forward in toe–off state (i.e. when the foot lose its contact with the ground at the end of a stance motion), a compression spring shock absorber is installed at toe region (see Fig. 3(d)).

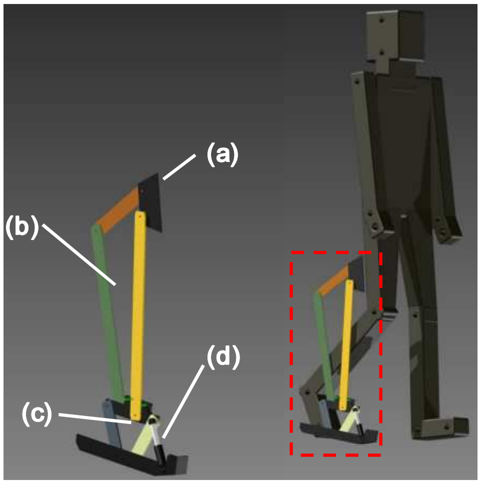


Fig. 3. The design of TAMC1

(a) Thigh seat (b) Closed–loop five bar linkages
(c) Closed–loop four bar linkages (d) Shock absorber

B. Prototype Fabrication

Based on the design, a prototype was built to verify the performance. For the material selection, an aluminium alloy was selected for its low cost, relatively easy manufacturability, light weight and moderate strength. The prototype is shown in Fig. 4. In order to constrain the ROM (range of motion) of upper link (see Fig. 4(b)), two limiters (see Fig. 4(g)) and a tension spring (see Fig. 4(c)) are installed. Each link has multiple joint holes for adjusting the size of device.

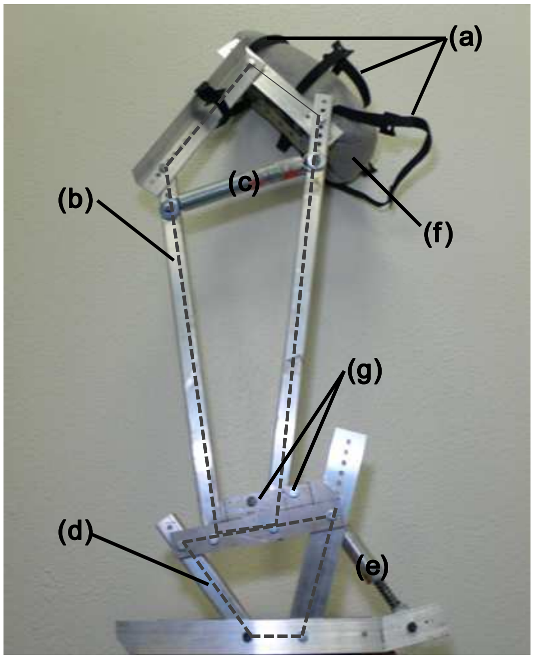


Fig. 4. The prototype of TAMC1

(a) Thigh straps (b) Closed–loop five bar linkages
(c) Tension spring (d) Closed–loop four bar linkages
(e) Shock absorber (f) Thigh seat
(g) Upper link limiters

C. Experiment

The fabricated prototype of TAMC1 was worn by a volunteered subject weighed 155 lb. Each link length is adjusted to fit the crutch to the subject. During the stand still posture, the TAMC1 could sustain the body weight. However, the attachment mechanism was not secure enough to allow any locomotion.

V. FINAL DESIGN CONCEPT

From the tests on the preliminary design, the final design is determined as a serial chain with a thigh cuff attachment. This design is inspired by the passive walker, which was first introduced by McGeer [21]. The passive walking device is a human–like walking device which is driven by gravity only. Due to its anthropomorphic shape, it is able to approximate the natural human walking very closely. Also, the serial chain can be fabricated in a light weight due to its simple design.

A. Design and Prototype

Based on the passive walking device design, several factors are modified according to the design objectives and specifications (see Section II) as follows:

1) For a secure attachment, thigh cuff is adopted and the subject's thigh is regarded as the first link of the chain. This also allows controllable and more stable locomotion by incorporating the subject's hip joint motion into the system.
2) Since the knee joint requires two different movements according to the walking phase (i.e. knee lock for a stance phase and free swing for a swing phase), an additional locking mechanism is adopted.
3) In order to guarantee a firm stability during a stance, a high stiffness ankle and foot mechanism is applied.

The final design of crutch, TAMC2, is shown in the Fig. 5. The entire thigh segment is covered by the cuff (see Fig. 5(a)). The thigh cuff consists of two parts and can be easily assembled by Velcro straps (i.e. usability– easy put on and off). Also for a secure attachment, each part has S–shaped interface (see the highlighted line in Fig. 5(a)) which can prevent relative translation. For the safety from unintended knee buckling, a prosthetic knee donated by a local prostheses and orthoses place is utilized (see Fig. 5(c)). The knee joint can be locked in any angle when a vertical force is applied (i.e. whenever the subject collapses). Also it is designed to be easily flexed with a small amount of force at the toe off posture and to be retracted when the foot loses its contact with the ground. The first link, thigh cuff, and the second link, knee prosthesis, are joined by a steel bracket to ensure the structural strength.

The fabricated prototype is shown in Fig. 6(a) and an enlarged picture of the prosthetic knee joint is represented in Fig. 6(b). Thermoplastic is utilized for the

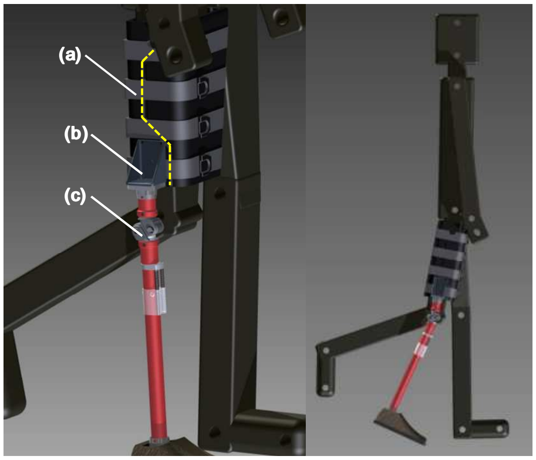


Fig. 5. The design of TAMC2

(a) Thigh cuff (b) Joining bracket
(c) Prosthetic knee joint

(a) Final prototype (b) Prosthetic knee joint
Fig. 6. The final prototype, TAMC2

thigh cuff. For stable and stiff ankle–foot mechanism, a frame of bike seat is adopted. A tracking shoe is worn on the structure for shock absorption and natural look.

B. Testing the Foot Trajectory of the Final Design in the Vicinity of the Specified Positions

As mentioned in the Section III, the two task positions of a gait cycle are determined as heel strike and toe off. In order to test the foot trajectory of the final design, the kinematic specifications at two task points on the reference trajectory are derived. Fig. 7 shows a foot trajectory of one cycle of normal treadmill walking which is a part of Fig. 2(b).

The kinematic specifications obtained from the Fig. 7 are applied to the design equations to test the smooth movement of the RR chain throughout the task. Fig. 8 shows the RR chain going through the specified task, consisting of two positions, two velocities and one acceleration, defined in the first position.

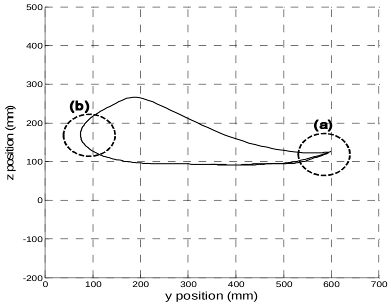


Fig. 7. The reference foot trajectory of one normal gait cycle
(a) Heel strike/first position (b) Toe off/second position

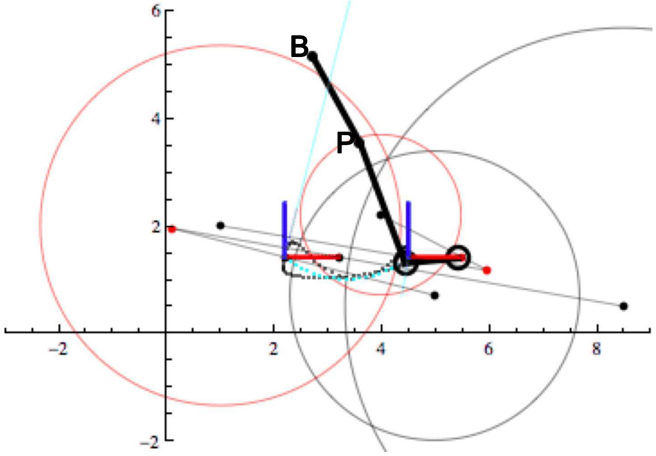
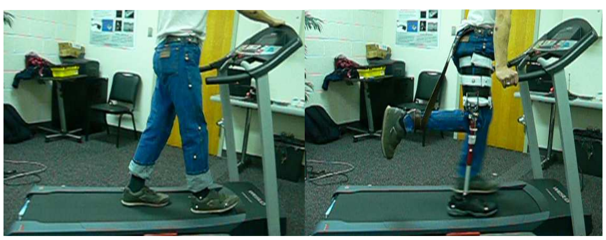


Fig. 8. The test result of the kinematic synthesis of RR chain **B** and **P** represent the fixed and moving pivot, respectively

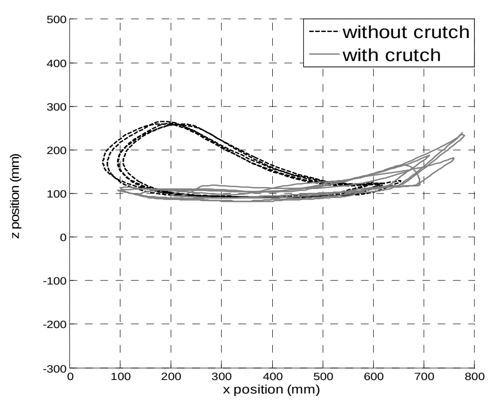
For the trajectory tests, equation (20) is used for finding locations of the fixed pivot (i.e. hip joint) and the moving pivot (i.e. knee joint) with respect to a fixed frame, located at the hip joint. The kinematic specifications at task points assist in shaping the foot trajectory. The trajectory shown in Fig. 8 represents the foot path after the contact and curvature constraints have been implemented into the task. Note that the tested trajectory with velocity and acceleration specifications is close to the desired one (i.e. '*teardrop shape*') in the vicinity of the two specified positions, heel strike and toe off.

VI. TREADMILL WALKING EXPERIMENT

The volunteered human subject performed treadmill walking experiments with and without the TAMC2 prototype. In order to emulate the lower leg injured condition, the subject tied one's leg to the waist (see the right side figure in Fig. 9(a)). To test comfort, equation (20) was used. According to the proper position of fixed frame from the kinematic synthesis, attachment location of thigh cuff was determined. The speed of treadmill was set as 2 mph for both experiments and the end–foot (i.e. ankle marker) positions are recorded by the 3D Motion Capture System. The acquired trajectories are compared as shown in Fig. 9(b). Unlike the normal trajectory or the tested trajectory (see Fig. 8), the result with crutch was formed flatter than the desired one. The main reason for that is the self–retracting mechanism of the prosthetic knee joint. Because of its stiffness, the knee joint extended faster than the hip flexion. This phenomenon eliminates and flattens the swing phase trajectories.



(a) Pictures of experiments (left: without crutch, right: with crutch)



(b) Foot trajectory compairment

Fig. 9. Treadmill walking experiments and their results

VII. SUMMARY AND FUTURE WORKS

In this paper, the disadvantages of conventional designs, underarm and forearm crutches, are studied. In order to overcome them, required design objectives and specifications are setup. Closed–loop linkages design is considered as a preliminary design. From the preliminary design tests, the final design was determined as a serial chain inspired by McGeer's [21] passive walker. To fulfill the two main design objectives, natural walking motion and comfort, a kinematic synthesis of RR chain was applied. The kinematic specifications (i.e. position, velocity and acceleration compatible with contact and curvature constraint between the foot and the ground) at two important gait events (i.e. heel strike and toe off) are derived from the normal foot trajectory. By imposing the obtained kinematic specifications to the synthesis design equations, both factors, the locations of the fixed and moving pivots, as well as the foot trajectory have been considered, i.e. comfort and natural gait motion.

As future works, the knee joint of the crutch will be modified to more closely resemble the natural human walking motion. In order to maintain the knee retracting function while more natural knee flexion is required, a semi–active actuator such as an electromagnet can be adopted.

ACKNOWLEDGMENT

The authors greatly acknowledge the support of NASA TSGC design challenge program, NASA mentor, Regnald Berka, Central Texas Orthotics and Prosthetics, as well as the members of the design team, Andra Arnold, John Bonnette, Tom Carney and Daryll Zalesak.

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