An Evaluation of Crutch Energetics using standard and “hands-free” crutches.
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Abstract
Crutches are commonly used to assist ambulation among disabled people, particularly those with below the knee injuries. Previous studies have compared mechanical energetics of various standard crutches with normal walking in order to determine energy expenditure variations. The primary purpose of this study was to compare the two-dimensional mechanical energetics of normal walking, swing-through gait with underarm crutches and a novel “hands-free” crutch. The “hands-free” crutch provides patients with non-weight bearing lower leg injuries an alternative to standard crutches while allowing full use of their arms and hands. Potential, rotational and translational energies in the anterior-posterior plane were considered. Thus, the secondary purpose was to evaluate the adequacy of two-dimensional mechanical energetic analysis using a eleven-segment model.

Preliminary results indicate possible trends between the energy costs of the three modes. However, further studies are required to validate these trends. It should also be noted that potential, rotational and translational energy paths differed significantly among the three modes. It is likely that the novel “hands-free” crutch is more efficient than standard underarm crutches, although further analysis with increased sample size is necessary.

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Introduction
Mechanical walking aids have been extensively used as locomotive assistance for many centuries. Particularly, crutches are most commonly used to reduce or eliminate weight bearing on a lower limb. The form of crutch locomotion adopted by most crutch users is known as swing-through gait since the subject moves forward by altering support on one or both feet and on the crutches (1). However, one major disadvantage is that this form of assisted locomotion requires a much greater energy demand than normal walking. Another disadvantage of this transport mode is that it inappracates any use of the upper limbs or hands.

Recently, an alternative for below knee, non-weight bearing injuries has arrived on the market. The ‘hands-free’ crutch was developed to enable the subject to freely utilize their upper limbs and hands while reducing or eliminating weight bearing on the injured limb. Thus far, the product has caught the attention of the medical industry as well as the general public.

It is quite likely that this aid could eventually become a standard locomotive aid routinely prescribed by physicians. The novelty and increasing popularity has prompted this investigation as to its mechanical energy cost compared to standard under-arm, crutches.

Several studies have previously reported that the ratio of energy cost in swing-through gait compared to normal walking ranges anywhere from 1.3-6 (1). It should be noted that the fairly large discrepancy is likely due to the different experimental procedures used in each study, type, size and shape of ambulation surface, the age, skill and health of the subjects and the velocities studied (1). Though many investigations have studied the energy costs (both metabolic and mechanical) of swing-through gait, there is no published literature on any type of device resembling the ‘hands-free’ crutch. The investigators feel that a mechanical energy analysis of this device would be of great interest to medical professionals as well as patients using the device.
Mechanical energy is a single parameter that employs information relating mass, moment of inertia, linear velocity, angular velocity, and force. Many different approaches have been taken to evaluate these parameters and combine them into a mechanical energy model. Various methods have been previously described (2,3,4,1) and are continuously debated (see discussion). Further, the debate surrounding this issue continues to interfere with the development of a well-understood 'gold standard' model.

The purpose of this investigation was to compare the total mechanical energy during one complete stride using two different locomotive aids, a 'hands-free' crutch and standard swing-through gait crutches with normal walking. The method of instantaneous energy analysis was used as described by Winter et al. (3,4), with some minor modifications. Due to constraints beyond the investigators control, a very basic 2 dimensional analysis was performed. Although the validity of this analysis could be debated, results from a pole vault study by Schade et al. (5) indicate that energy measurements with 2-d and 3-d techniques do not significantly differ (see discussion).

Methods

Subjects and Procedure

Three male subjects, ranging in age from 21-24 were used for this study, with no neural, muscular or skeletal abnormalities. The subjects were asked to walk normally, with swing-through gait crutches, and a 'hands-free' crutch for several trials across test area. Each subject was allowed time to familiarize themselves with each locomotive mode before data acquisition began. Subjects were filmed in the anterior-posterior plane for one complete stride (heel strike to heel strike) for each transport aid at a comfortable, 'normal' pace. The immobilized limb was consistent for all subjects.

The subjects were video taped with a 60 Hz recorder from the same side of the body for each trial. Markers were placed on both sides at the wrist, elbow, shoulder, hip, ear, knee, ankle, toe and heel. The video was manually digitized using the Peak 5 Performance analysis system. Raw data was filtered using a two-pass, second order Butterworth filter with a cutoff frequency of 6 Hz. This was required in order to reduce the 'noise' in the signal. Further smoothing of the filtered data was achieved by using a best-fit polynomial of 6th degree. In order to eliminate discontinuities due to filtering, data points from the end of the cycle were added to the beginning, and data points from the beginning were added to the end. This created a more accurate representation of one complete gait cycle.

Measurement of Mechanical Energy

An eleven-segment model was used to calculate the different energies required for analysis. It should be noted that the head, trunk and neck (HTN) were grouped as a single segment. Marker positions, anthropometrical tables and subject height and weight were used to calculate the centre of mass, radius of gyration, inertial parameters, centre of mass, segment length, and change in angle of each body segment (3).

The laboratory coordinate system was set so that the direction of travel was the positive x-axis while movement from the floor upwards was the positive y-axis. The initial point of heel contact was considered to be the base value from which all potential energy was calculated.

Total mechanical energy of a subject at any point in time was calculated as the sum of the potential energy and kinetic energy of each body segment for each normalized time point. All calculations were computed based on the proximal joint for consistency. Kinetic energy was subdivided as rotational and translational energy. Hence, the total energy was determined by the following equation:
\[ E_{\text{total}} = \sum_{n=1}^{11} (PE + KE + RE) \]

**Equation 1: Total Energy**

Where \( n \) denotes each of the 11 segments and the rotational, potential and translational were calculated as follows:

\[ E_{\text{kinetic}} = \frac{1}{2}(i\omega^2) \]

**Equation 2: Rotational Energy**

\[ E_{\text{Potential}} = mgh \]

**Equation 3: Potential Energy**

\[ E_{\text{Kinetic}} = \frac{1}{2}(mv^2) \]

**Equation 4: Translational Energy**

Angular velocity (\( \omega \)) was calculated with the floor as reference. At any point in time, the angle (\( \theta \)) was determined from the following equation:

\[ \theta = \cos^{-1}\left(\frac{P_y - D_y}{L}\right) \]

**Equation 5: Angle Calculation**

Where \( P_y \) is the proximal position of the segment in the y direction, \( D_y \) is the distal position in the y direction and \( L \) is the length of the segment. The angular velocity for any point can be found by using the change in angle from the previous point and the change in time (for 60 Hz, each time interval equal to 1/60s):

\[ \omega = \frac{\Delta \theta}{\Delta t} \]

**Equation 6: Angular Velocity**

Moment of inertia (\( I \)) was calculated for all the individual segments using the parallel axis theorem in order to project rotation about the proximal joint:

\[ I = mr^2 + mx^2 \]

**Equation 7: Moment of Inertia**

Where \( m \) is the segment mass, \( r \) is the radius of gyration and \( x \) is the distance between the centre of mass and centre of rotation. The centre of rotation was always calculated at the proximally located joint.

Translational kinetic energy was calculated from the combination of vertical and horizontal centre of mass velocity:

\[ \sqrt{(CM_{x_2} - CM_{x_1})^2 + (CM_{y_2} - CM_{y_1})^2} \]

**Equation 8: Center of Mass Velocity**

Where CM denotes centre of mass, \( X/Y \) indicate the x and y coordinates, respectively, and the subscript indicates two consecutive time frames.

The data was time normalized so that 100% gait represented one complete gait cycle (heel strike to heel strike). Also, total body energy was expressed as percent of body weight (N) multiplied by the average leg length (LL in meters). This facilitated summing the total segmental energies and eliminated time and size variations among subjects and locomotive modes. This technique was previously used by Miller et al. (2). Energy efficiency was calculated by integrating the 6th degree polynomial for each curve. The area was taken with respect to the initial energy at heel strike.
Results

The results are shown graphically in Figures 1 and 2.

Figure 1: Individual Subject Data

Figure 2: Comparisons Among Subjects
Figure 1 depicts energy curves for each of the three subjects tested. Figure 2 shows a comparison of energy curves for each transport mode of all subjects.

Figure 3 is a graph of the energy efficiency calculated as a percent of normal walking. Figure 4 is an example of the various energies that compose the total body energy.

**Discussion**

Upon analysis of Figure 4, it is evident that the total energy is a result of the summation of the potential, rotational and translational energies. It is also quite apparent that the potential energy accounts for the majority of the total energy fluctuation observed, while the rotational and kinetic are relatively small in magnitude. This seems to be consistent with the results from previously published data (3, 1, 2). This would indicate that the 2d model might provide a close approximation for the total body mechanical energy. It should also be noted that the out-of-phase curves of the potential and kinetic energies indicate that the body acts similar to that of a simple pendulum during swing-through gait crutch walking.

Figure 2 shows the energy paths of each subject for each locomotive mode. It is quite apparent that each subject exhibited the characteristic double peak for the normal walking trial (3). This would indicate that the 2d instantaneous energy analysis demonstrates similar trends to those computed with more complex 3d procedures (2). This would suggest that a 2d analysis might be adequate for mechanical energy analysis of relatively simple movements.

The crutch data demonstrated a trend with an initial decrease in energy followed by an obvious increase towards the end of the stride. This seems to be consistent with previously published data at similar walking speeds (1). It is likely that this mid-cycle surge is due to the large increase in kinetic energy and potential energy at the end of the swing-through phase. The shape of the energy path for each subject was relatively consistent. Any deviations were likely due to inexperience or discomfort with swing through crutch walking.

The energy path for the novel ‘hands-free’ crutch exhibited some similarities,
although no significantly consistent trends were observed. From the data, it is likely that the end-cycle surge is due to the normal leg being moved forward, as heel strike of the mobile leg occurred before crutch contact. Since the crutch-thigh segment behaves as a single straight piece, the subject must increase the segment energy in order to raise the leg and clear the ground. It is also possible that the second hump is similar in nature to that of the walking curve since the second half of gait cycle with the crutch closely resembles that of walking.

Figures 1 and 3 show individual subject data. Total energy efficiency was determined from each curve in Figure 1. The integration of the trendline equation with respect to the initial point of heel strike was calculated and plotted as a bar chart in Figure 3. This technique was used since the data had been already normalized for time, body weight and stride length (represented by leg length). The area results indicate a similar trend among two of the three subjects where the crutch demonstrated the highest energy inefficiency followed by the ‘hands-free’ crutch, compared to normal walking. This trend was expected since previous swing-through crutch studies have indicated significant increases in energy compared to normal walking (1). Further, the increasing demand of the crutch, comfort level noted by the subjects and functional design would suggest some energetic advantage for the ‘hands-free’ crutch.

However, although this trend was generally expected, one subject’s data failed to follow. In fact, according to the results, this subject was more efficient with both crutch types than when walking normally. This obvious error is likely due to the manual digitizing of the video or inadequate representation of the data by the trendline, both of which could result in erroneous calculations.

It is also interesting to note that one subject demonstrated approximately 5 times more energy expenditure with the swing-through crutch than normal walking. This result could be due to the obvious discomfort demonstrated by the subject and a proportional inefficiency or could be due to manual error. The subjects efficiency was still in the range indicated by Thys et al. (1).

Limitations of the study

The results obtained thus far should be considered preliminary, as many limitations were present. For example, the instantaneous energy model does not account for isometric muscle contractions and any resulting energy losses. Therefore, a complete analysis would have to include EMG data. Further, total body mechanical energy is only significant when compared to metabolic energy demands such as cardiac output, VO₂ max and heart rate. The reason for this is that ratio is often described as the ratio of mechanical to metabolic energy. Thus, this ratio describes how much energy is actually used to advance the body during locomotion. Many investigations have previously shown the importance of this relationship (3,1,2,5).

The obvious criticism of the results would be that a 2d analysis does not properly model a 3d movement. Thus, it is fairly intuitive that a 2d model would have significant validity issues when dealing with any action with definitive medial-lateral movement. However, one could hypothesize that because of the symmetrical nature of ideal normal walking, there would be very little movement in this plane. When compared to previously recorded data, it is evident that the results indicate the presence of the characteristic shape and magnitude of the double peak expected in normal walking (3). This indicates that any movement in the medial-lateral plane, especially for walking can be ignored for energy calculations. Further, a study conducted by Schade et al. (5) revealed that the percentile differences for selected energetic parameters were less than 1% for the 2d approach compared to the 3d approach. The fact that the highly unsymmetrical nature of pole vaulting is
does not significantly affect the model used, proposes that any medial-lateral movement in our locomotive modes could also be neglected.

Another debatable issue arises with the procedure used to determine mechanical energies. The deficiency of an agreed upon and well-understood ‘gold standard’ has been the subject of extensive debate in the literature. Particularly, the two procedures in question are calculation of mechanical energy using the centre-of-mass approach vs. the total body approach (5).

The centre of mass approach is based on the idea that total mechanical work is the sum of the internal and external work (1). Thys et al. (1) used this method where external work was defined as the work necessary to move the body centre of gravity and internal work was defined as the work necessary to move the body segments relative to the body centre of gravity (6). The upper and lower limits to the total work were determined by including or excluding all possible energy transfers between segments of each limb, between the limbs themselves and between the body centre of mass and the limbs (7). The major assumption of this model is that the work performed by the body is not equal to the work performed on the body. Thus, the main argument against this model is that simultaneous increases and decreases in oppositely moving segments would not be accounted for. This model also assumes that the body’s centre of mass represents the energy changes in all the segments. Therefore, this approach does not account for the energy losses from energy absorption and generation at different joints. This method has been highly disputed in the literature (4,5,3,1,6).

Winter et al. (3) among others have previously used the instantaneous energy technique. This method requires treating the total body energy at any instant as the sum of the energy of each segment. Segmental energy is comprised of potential, kinetic and rotational energies. This technique recognizes both the conservation of energy within each segment as well as the energy transfer between adjacent segments (3). However, this calculation is known to underestimate the simultaneous energy generation and absorption at different joints. The total body work will usually produce a low estimate of the positive and negative work by the human motor system from this method (3). Schade et al. (5) have also shown similar results in their study of pole vault energetics.

Finally, the most obvious limitation was the small sample size. Unfortunately, this parameter was beyond the control of the investigators due to course requirements, time period and extensive digitizing requirements.

In conclusion, the results obtained from the study, although preliminary, indicate exciting potentials for further study of the ‘hands-free’ device. Without doubt, a more comprehensive, interdisciplinary study is necessary to assess parameters such as metabolic efficiency, balance, load distribution and fracture healing. The resulting implications of these future investigations could provide a valuable alternative for below knee injuries.

References